SEATED POSTURAL CONTROL IN SPINAL CORD INJURY

BY

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DISSERTATION

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Abstract

This is a cross-sectional study for seated postural control in spinal cord injury. The general goal of the present study was to further understand seated postural control of individuals with SCI in various different conditions (e.g. manipulations of vision, proprioception, and task constraints). The overall purpose was to test the predictions of the loss of complexity hypothesis and the loss of adaptability hypothesis on seated postural control of persons with and without SCI. The general hypothesis was that seated postural control of individuals with SCI would be influenced by SCI injury level, sensory information (vision and proprioception) and task constraints. Three experiments were conducted to test the hypothesis. These experiments consisted of a series of manipulations of a sensory information (proprioception) and task demands. The results of the present research project leads to the general conclusion that even though persons with SCI had reduced postural control which were indexed by increased sway area, reduced dynamic structure, and shorter VTC, they actively use additional sensory input and develop different postural strategies to maintain stability.

To my family in God.

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Chapter 1

Introduction

1.1 Background

There are over 250,000 Americans living with a spinal cord injury (SCI) and approximately 11,000 new injuries occur each year (Center, 2004). Also it was reported that at least 1.5% to 3% of major trauma victims are due to cervical spinal cord injury and most of these cases occurs between 15 and 35 years of age (Urdaneta, Layon, Guiot, Mendel, & Kirby, 2003). Spinal cord injury leads to a change in mobility and social interaction.

It is well known that there is a functional limitation in mobility with SCI that negatively impacts quality of life. Sitting is one of the most frequent and fundamental postures of daily life (Shirado, Kawase, Minami, & Strax, 2004). Persons with SCI face significant challenges in maintaining a seated posture. Postural control, the ability to maintain the center of mass of the body within specific boundaries of space, is achieved by integrating multiple sources of sensory input such as vision, vestibular sense and proprioception and appropriate motor output based on these inputs (Jeka, Easton, Bentzen, & Lackner, 1996; Hong, Manor, & Li, 2007). Dysfunction in seated postural control in persons with SCI stems from impaired neural control of the involved musculature as well as a decrease in the sensory information being transmitted to the brain.

Posturography or stabilometry in which force platforms measure fluctuations in the application of the point of ground reaction force has been one of the most widely applied techniques to access standing postural control (Winter, 2005). In traditional posturograpy, COP parameters describe amplitude of sway such as mediolateral (ML) and anteroposterior

(AP) displacement and velocity of these parameters have been used for decades (Cavanaugh, Guskiewicz, Giuliani, et al., 2005). Mean, standard deviation (SD), and coefficient of variance (CV) from multiple trials of COP produced by standing body sway have been used as linear descriptors to quantify the motor variability (Prieto, Myklebust, Hoffmann, Lovett, & Myklebust, 1996). However, these measures do not account for the time-evolving properties of postural control (Newell, Emmerik, Lee, & Sprague, 1993; Zatsiorsky & Duarte, 1999; Collins & De Luca, 1993; Collins, De Luca, Burrows, & Lipsitz, 1995).

It is well known that physiological function, such as postural control results from the complex interaction of multiple control mechanisms operating over unique timescales (Lipsitz & Goldberger, 1992). This interplay of control mechanisms is the driving factor underlying the time-evolving dynamics of physiological output. It is maintained that changes in physiological function resulting from aging, disease or injury lead to decrease in the number of and interaction between control mechanisms and that this decrease in control mechanisms is characterized by an increase in the dynamic structure of physiologic output, e.g. loss of complexity (Lipsitz & Goldberger, 1992; Goldberger, Peng, & Lipsitz, 2002). Moreover, Lipsitz and Goldberger (1992); Goldberger et al. (2002) have proposed that the loss of complexity (e.g. the observed output signal becomes more regular and less complex) results in functional decline.

In addition to being of theoretical interest, the structure of standing postural sway time series has also been found to be more sensitive than traditional linear parameters when distinguishing various populations (Cavanaugh, Guskiewicz, & Stergiou, 2005; Newell et al., 1993; Sosnoff, Broglio, Shin, & Ferrara, 2011). Nevertheless, previous research examining seated postural control in SCI compared sway ranges between SCI population and able-bodied controls (Seelen, Potten, Huson, Spaans, & Reulen, 1997; Seelen, Janssen-Potten, & Adam, 2001; Shirado et al., 2004). Therefore, there is no information concerning the time evolving structure of seated postural sway in persons with SCI.

Despite the large amount of empirical evidence in support of the loss of complexity

hypothesis (Duarte & Sternad, 2008; Schmit et al., 2006; Thurner, Mittermaier, Hanel, & Ehrenberger, 2000), there is growing criticism. For example, most work examining this theoretical framework has assumed decrease in control processes (e.g. number of DoF) as a function of advanced age and disease. The actual number of control processes is rarely defined. An examination of persons with SCI would allow for a quantification of the control processes involved due to their injury.

Although it is logical to assume that persons with SCI will demonstrate a decrease in complexity in seated postural control compared to able-bodied controls, there is growing evidence that physiological complexity does not simply decrease with advanced age or disease (Newell, Vaillancourt, & Sosnoff, 2006). But rather, changes in complexity are driven in part by the interaction of the organism and task demands. This proposition forms the tenets of the loss of adaptability theory. This theory maintains that the complexity of the system can be increase or decrease dependent on the tasks being performed (Sosnoff, Valantine, & Newell, 2009; Sosnoff & Newell, 2008).

Finally, the general goal of the present study was to further understand seated postural control of individuals with SCI in various different conditions (e.g. manipulations of vision, proprioception, and task constraints). The overall purpose was to test the predictions of the loss of complexity hypothesis and the loss of adaptability hypothesis on seated postural control of persons with and without SCI. The general hypothesis was that seated postural control of individuals with SCI would be influenced by SCI level, sensory information (vision and proprioception) and task constraints.

Three experiments were conducted to test the hypothesis. These experiments consisted of a series of manipulations of a sensory information (proprioception) and task demands. They are briefly outlined below.

The investigation outlined in Chapter 3 determined if spinal cord injury influences complexity of seated postural control according to the predictions of loss of adaptability hypothesis. In accordance with the loss of adaptability theory, it was predicted that persons with

SCI will have bidirectional complexity driven by the task constraint (i.e. still sitting vs. voluntary rocking). It was expected that participants would demonstrate higher complexity in body rocking, but lower complexity in quiet sitting. In order to address this prediction, 10 persons with SCI and 10 age, gender and sitting height matched controls without SCI conducted the voluntary body rocking and sitting still.

Chapter 4 presented an investigation on the influence of additive proprioceptive information. The second aim of this project was to determine if light touch would differently affect complexity and variability of postural control in populations with SCI and without SCI during quiet sitting. It is predicted that light touch would increase the complexity in postural control of persons with higher SCI injury. In order to address this prediction eight persons with SCI and eight age, gender, and sitting height matched controls conducted sitting still with three different touching condition (i.e. no touch, heavy touch (over 1N) and light touch(under 1N)) under two different sitting surface condition (i.e. stable condition(without flexible disk) and unstable condition (with flexible disk)).

Chapter 5 outlined an investigation examining if decreased complexity would be related to greater functional impairment defined as decreased time to contact to the stability boundary. The fourth aim of this project was to determine if SCI influences time to instability in sitting postural control. It was predicted that postural control in sitting of persons with higher SCI lesion would have shorter virtual time to contact to the functional boundary (VTC) and their VTC would be less complex. To test the predictions of the loss of complexity, most of the work examining this theoretical framework has assumed with aging there is a decrease in DoF between populations (Lipsitz & Goldberger, 1992). Even though the actual number of DoF was rarely defined in the previous studies, in a population such as spinal cord injury the number of DoF could be defined by injury. We believed that defining the actual number of control processes (e.g number of DoF) by the injury level is a more effective way to examine the hypothesis of study. Therefore, we divided participants into three groups (high spinal cord injured (HI), low spinal cord injured (LI) and non-spinal cord

injured groups (NI)). In order to address this prediction, 7 persons with high SCI, 11 persons with low SCI and 18 age and gender matched controls conducted lean forward, backward, laterally and diagonally pivoting at the hip joint in the sitting posture to determine their stability boundary (functional boundary). With combination of stability boundary and the still sitting data, virtual time to contact stability boundary (VTC) was calculated.

Chapter 2

Review of literature

2.1 Spinal cord injury

There are over 250,000 Americans living with a spinal cord injury (SCI) and approximately 11,000 new injuries occur each year (Center, 2004). Also, it was reported that at least 1.5% to 3% of major trauma victim is cervical spinal cord injury and most of these cases occurs between 15 and 35 years (Urdaneta et al., 2003). Spinal cord injury leads to a change in mobility and social interaction.

To individuals with spinal cord injury, sitting is one of the most frequent and fundamental postures of daily life (Shirado et al., 2004). Seated postural control is a not simple passive alignment of body segments, but rather a complex task which requires multi-sensory information such as visual, vestibular and proprioception (Gagnon, Vincent, & Noreau, 2005). Persons with spinal cord injury usually have to maintain a sitting posture for an extend time during physical activities, and it is maintained that poor postural control in person with SCI will impair their ability to perform functional activities (Petrofsky, Cuneo, Lee, Johnson, & Lohman, 2006).

2.2 Physiological basis of spinal cord injury

2.2.1 Consequence of spinal cord injury

There are two categories of spinal cord injury; one is the destruction of the intraspinal neuronal apparatus and the other is a deterioration of neural conduction pathways (Kandel, Schwartz, & Jessell, 2000; Latash, 2008). The intraspinal neuronal apparatus is composed of interneurons and motoneurons. The damage to the alpha-motor neurons, which are located within the gray matter, make normal voluntary control of skeletal muscle impossible, and the destruction of the interneuron within neural structure may impair motor coordination.

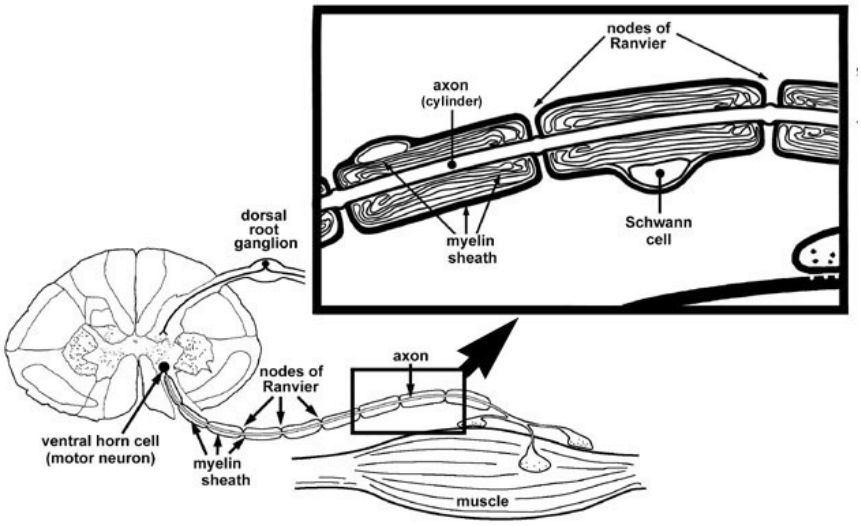

Figure 2.1: Neural structure in spinal cord; adapted from Harting et al. (1997).

Figure 2.1 shows the neural structure in the spinal cord. The neural structure in the spinal cord contributes to voluntary motor action of skeletal muscle. It is believed that spinal mediated control stems from central pattern generators (CPGs) as well as reflex pathways (Latash, 2008).

2.2.2 Spinal cord injury and functional limitation

It is well known that there is a functional limitation in mobility with SCI that negatively impacts quality of life. Generally, maintaining seated balance of a non-SCI population is an automated process in which little attention is required, whereas this process in people suffering from SCI is damaged (Seelen et al., 1997; Seelen, Potten, & Pons, 1999). Spinal cord injury largely can be divided into tetraplegia and paralegia by the severity and level of injury. Tetraplegia or quadriplegia is "a general term of describing the condition of an individual having a spinal cord injury at a level from C1 to T1" (Bromley, 1976).

Normally, the individuals with tetraplegia have a loss of feeling and movement in their neck, shoulder, arms and/or upper chest. Paraplegia is "the term describing the condition of an individual who has lost feeling and/or is not able to move the below injured parts of his/her body" (Kirshblum & O'Connor, 2000). The body parts that may be loss in function are the chest, stomach, hips, legs and feet.

With regard to loss of function in seated postural control, the level of compensation for the loss of function in postural control is influenced by the SCI level. If the SCI level increases, the required compensation levels also increase. However, it is impossible to fully compensate for loss of balance control because as the SCI level increases the possibilities of the compensation decrease. As SCI level increases, impairment is not only found in the motor system, but also exterocepsis (stimuli from skin sensors) and propriocepsis (stimuli from sensors in the muscle spindle, tendons and joints) in thoracic SCI which are essential for adequate postural control (Massion, 1992; Massion & Dufosse, 1988; Seelen et al., 2001). Traditionally, the purpose of rehabilitation of SCI patients is not full recovery from the lesion but to tune new acquisition of skills (Geurts, Mulder, Rijken, & Nienhuis, 1991) or the re-automatization of postural control and (focal) movement which requires the ability of a participant to anticipate forthcoming balance-perturbing events and postural changes (Mulder, 1991; Massion, 1992).

Seelen et al. (2001) demonstrated that for individuals with high SCI (injured from T2 to

T8) compensation for the loss of postural muscle function by using more portions of non-postural muscles is essential to erect the upper body (Seelen et al., 2001). It was reported that damaged function of the erector spinae (ES) in individuals with SCI causes a smaller center of pressure shifting during sitting (Seelen et al., 1997, 2001). Also, results about motor preparation of the SCI population verified by Seelen and Vuurman (1991) showed that during the perturbation by upper extremity reaching tasks, high thoracic SCI and low thoracic participants utilize different postural control strategies. Seelen and colleagues showed that high SCI participants show shorter reaction time in reaching tasks (Seelen et al., 2001). Reft and Hasan also reported that in a target reaching task, deficits in trunk control were related with reduced peak velocity of the hand and differentiate the path of the hand (Reft & Hasan, 2002). However, Jacquier-Bret, Rezzoug, Vallier, Tournebise, and Gorce (2009) report that C6-C7 quadriplegic subjects still have flexible joint configurations during a reaching movement in that they show "arm compensatory synergy" by showing similar features of the variance structure with Non-SCI subjects (Jacquier-Bret et al., 2009).

2.3 Theoretical perspectives on human postural control and movement variability and complexity in dynamic system approach

2.3.1 Degree of freedom problem

There are two theoretical viewpoints concerning how posture is controlled. According to the traditional view, postural responses are seen as innate reflexive action governed by the spinal mechanisms. However, from a dynamic perspective, postural control is considered an emergent skill occurring as a result of interacting with the environment in the process of solving the degree of freedom problem (Harbourne & Stergiou, 2003) whereas an information

processing approach to motor control, which views the human as a learning machine, give rise to the idea that rules for movement execution are collected in motor programs. Bernstein proposed that the problem of controlling movement is too complex to be explicable in terms of motor programs (Todorov, 2004). Bernstein questioned where a person could store the amount of information needed to produce the variety of movement we can see in everyday activities. This theorization lead to the formulation of the Bernstein problem: What is actually being controlled within the human system that results in a movement? (Bernstein, 1967). When we describe the motion in a space, the minimal number of descriptors required to characterize the object's position in space is the degree of freedom (DOF) in mechanical meaning. For example, motion of a rigid body in three-dimensional space has six DOF which can be described by translation in three perpendicular axes (x, y, z) and rotations about three perpendicular axes (pitch, yaw, roll). Another example is that the arm has seven DOF; three at the shoulder, two at the elbow, and two at the wrist. Actually, human body has more DOF than the number of joints necessary to conduct any tasks such as transporting body or reaching and grasping the objects in spatial positioning and orientation of hand on the object. In voluntary actions of humans, redundancy of DOF has become a central issue in the study of motor control (Latash, Scholz, & Schöner, 2007).

It is maintained that humans solve "Bernstein's degree of freedom problem" by learning how the body's dynamics interrelate with the demands of movement tasks. They exploit the movement-associated torques or interaction torques that bodies produce when they move. In the approach to solve the degree of freedom problem, Bernstein argued that the kinematic and dynamic aspects of movements, and the functional expediencies of these aspects were grossly undervalued in accounts of coordination (Turvey, 1990). He ruled out any straightforward, unambiguous relation between the nerve impulses innervating movements and the movements themselves. In doing so, he underscored the essential formative and steering roles of the information available to perceptual systems. However, Turvey (1990) argues that with his action plans concept, the referents appear to be relations among properties

that are relatively few in number, realizable in all body segments, and capable of generating many different motions. Bernstein suggested that the numerous mechanical DOF could be reduced through muscle linkages (or synergies, or coordinative structures). He considers movements not a unit of coordination but a unit in coordination. His attempt to solve all of coordination with general laws made him adopt intensions, plans goals, and so on into general laws. He thought the information in the ecological approach is related with specificity between the structured energy distributions to a perceptual system and the environmental and movement properties are casually responsible for the structure (Turvey, 1990). To make us understand coordination in terms of general principles, he suggested two methodological features. One is the understanding of "nonmonotonic patterns" of rhythmic movement in the non-linear dynamic approach or ecological approach. Another is that extension of measurable quantities. He focused not on traditional psychological methods such as reaction times and error, but on nonlinear dynamics of physical biology and information analyses of ecological psychology. All specific examples such as assembling many varied micro components into rhythmic units, investigations of mechanical oscillators with nonlinear stiffness, and oscillators or orbital attractors with balanced energy losses and gains support his argument strongly (Turvey, 1990).

Especially, in Turvey's approach, he knew that the central nervous system (CNS) cannot find a unique solution for the problem of kinematic redundancy by eliminating redundant DOF, but rather used the apparently redundant set of joints to ensure more accurate performance of the tasks. Bernstein's issue of degree of freedom problem is understood as multiple joint and muscle solutions for a given movement. The merit of this approach is that perceptual system, and the environmental and movement structure interact simultaneously in a task based on a few general laws. The idea is exceedingly reasonable and may be accepted with strong evidence. For example, changes in muscle activations and muscle torque after SCI such as cerebral vascular accidents and cerebellar damage could cause different coordination of movement shown as inaccurate and curved finger paths, mover variability of endpoint

positions, and multi-peaked fingertip velocity profiles (Bastian, Zackowski, & Thach, 2000; Beer, Dewald, & Rymer, 2000; Topka, Konczak, Schneider, Boose, & Dichgans, 1998).

2.3.2 Complexity

It is well known that physiological functioning results from a complex interaction of multiple control mechanisms (Lipsitz & Goldberger, 1992). Changes to physiological functions caused by aging, disease or external stress lead to the alteration of dynamic structure in the physiologic processes. This view of "loss of complexity theory", suggests that as control processes decline (e.g. dynamic DOF) the output signal becomes more regular and less complex, resulting in functional decline.

However, several recent experimental studies on human movement revealed that the notion of unidirectional change (e.g decrease) in complexity with aging and disease is too narrowly generalized. Vaillancourt and Newell (2002) argued that there can be both an increase and decrease of complexity according to the confluence of constraints in action.

As a contradiction to the evidence of a general decrease in complexity with aging, Vaillancourt and Newell (2002); Vaillancourt, Sosnoff, and Newell (2004) asserted that complexity should be changeable according to the task constraints. They showed that for a task where participants maintained a constant level of isometric force, complexity decreased with aging (Vaillancourt et al., 2004). However when tracking a sinusoidal wave of the same participants by varying isometric force, complexity increased with aging (See Figure 2.2). Their results implied that there will be bidirectional complexity changing by task constraints even with aging and diseases in postural control. Yates (2002) postulated that in the local structural regions or local dynamic signatures, complexity with aging could increase or decrease, but in the global perspective across whole anatomic and physiological aspects, loss of complexity with aging could be acceptable.

In the physiology and the motor control research, quantification of changes of physiology function is a key question. "Fractals" and "chaos" are two terms which are used in nonlinear

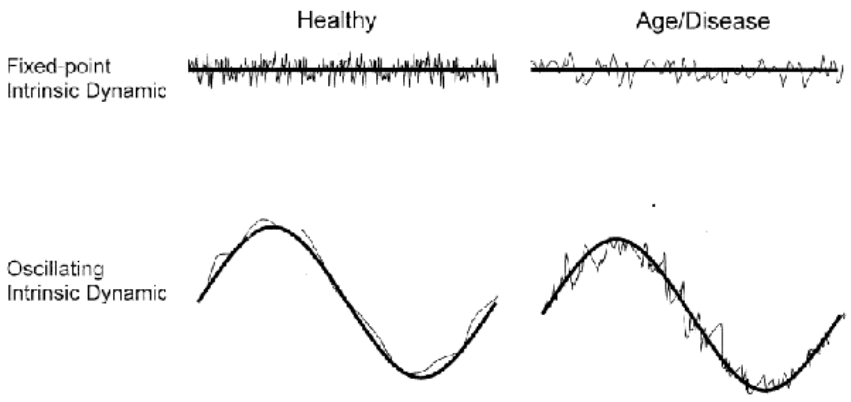

Figure 2.2: Schematic illustrating the fixed-point and rhythmical intrinsic dynamic posture. The fixed-point intrinsic dynamic (top) and rhythmical intrinsic dynamic (bottom); adapted from Vaillancourt and Newell (2002, p. 8).

dynamics to quantify dynamic structure in the complexity system (Liebovitch & Scheurle, 2000; Lipsitz & Goldberger, 1992). The term fractal relates to a structural feature or property having self- similarity in which small scale structure consist of the resembling the whole even though it is irregular pattern. Chaos is a description of unpredictable, complex behavior which is "noisy looking" with variable fluctuations.

Nonlinear dynamics analyses are considered as a useful tool to unveil the dynamic structure of a time (Lipsitz & Goldberger, 1992). To adequately capture nonlinear features, measuring fractal dimension and complexity of nonlinear system are often used. For example, approximate entropy (ApEn) is a method to determine complexity by quantification of the regularity or predictability of physiological data and Lyapunov exponent (LyE) is a measure of the instability of the system by quantifying the exponential separation of trajectories with time in state space. Correlation dimension (CoD) means measure of dimensionality of a dynamic system (Stergiou, 2004).

Measuring COP dynamics during quiet standing or sitting has been considered as a supporting experiment to examine the predictions of loss of complexity theory (Newell, 1998). The nonlinear characteristics of postural control justify the use of parameters describing the structure of motor complexity like approximate entropy, multiple scale entropy and Lyapunov

exponent (Madeleine, Nielsen, & Arendt-Nielsen, 2011). For example, it was reported that dynamic structures in COP fluctuations in given tasks comprise the nonlinear properties in the postural system and especially, in sitting postural control, can be an indicator of reorganization of degree of freedom, based on, mental retardation (Berkson & Andriacchi, 2000; Hong, Bodfish, & Newell, 2006), motor development or aging (Hong, James, & Newell, 2008; James, Hong, & Newell, 2009; Newell, 1998) and pathology (Lindenberger, Marsiske, & Baltes, 2000; Marchese, Bove, & Abbruzzese, 2003) in that it reflects the changes of number of freedom, i.e. the level of independence of muscles and joints coordination required for postural control (Newell, 1998; Hong et al., 2006). The level of dysfunction in maintaining posture during unsupported upright sitting is one of the highest predictors of functional outcome in various CNS pathologies, i.e. traumatic brain injury (Black, Sulayman, Baker, & Newcombe, 2000; Perlmutter, Lin, & Makhsous, 2010) and spinal cord injury (Bernard, Peruchon, Micallef, Hertog, & Rabischong, 1994; Seelen et al., 1997, 2001).

Most of previous experimental researches studies have examined the effect of aging and disease on the complexity of posture. Sitting still and body rocking tasks were used to investigate the complexity of posture. For example, Harbourne and Stergiou (2003) showed that infants' adaptation process in three different stages of sitting (entry, developing, and independent sitting). They argued that when complexity was changed by stage it can be evidence of self-organizing process of sitting. Hong et al. (2008) found that the COP path length and complexity decreased with age in children. They argued that the postural sway of younger children showed slower fluctuations that were more tightly coupled across the axes of motion when compared to that of adults. The finding in the sitting still task was consistent with loss of complexity hypothesis with aging and disease by Lipsitz and Goldberger (Hong et al., 2006). Recently studies using simulated body rocking motion have shown some theoretical advantage in using postural control study (Hong et al., 2008; James et al., 2009). Body rocking, a kind of rhythmic movement is useful to understand human movement because its mechanical resonance is more stable than any other movements at other frequencies.

Therefore, it can test how a person's ability tunes a motor system to the mechanical properties of the task and it is adequate to evaluate the control processing. For example,Hong et al. (2008) examined both microscopic entropy which were indexed by approximate entropy during the quiet still sitting and macroscopic entropy which were indexed by information entropy of relative phase values calculated by normalized COP position and velocity during body rocking task including a kind of rhythmical intrinsic dynamic properties between the stereotypic disordered group and their age matched controls. Their results showed the stereotypic disordered group has less microscopic complexity, i.e. lower approximate entropy value and less macroscopic entropy-more tightly coupled motion in quiet sitting. However, there were no differences in body rocking task which assess the rhythmical intrinsic dynamic with macroscopic level perspective to postural dynamic. It can be expected that the natural ability to tune movement in action to the properties of the task and environment may evaluate their injured level. In the adaptive feature of human behavior, their loss of neuromuscular function will make it changing the resonance mechanical to use the erect muscle differently.

2.4 Measurement of Postural control

Duarte and Freitas (2010) divided "posturography", a general term including all analysis for postural control, into global and structural analysis. Global analysis is a method to measure the size of oscillatory patterns both in the spatial and frequency domains whereas, the structural analysis quantify sub-unities or structures in the posture posturographic signals which is related with the motor control processes (Duarte & Freitas, 2010).

2.4.1 A global analysis: traditional methods to measure COP during postural control

A force platform is a common piece of equipment used to quantify postural sway in laboratory settings. It provides indirect measures of changes in postural sway by recording the ground-reaction forces projected from the body. From these ground reaction forces, center of pressure (COP) can be calculated. Center of pressure reflects the trajectory of center of mass and the amount of torque applied at the support surface (Palmieri, Ingersoll, Stone, & Krause, 2002).

Center of pressure (COP) measuring by using the force platform has been used as one of the most widely applied ways in accessing postural balance related with fall risk and medical assessment after injury (Piirtola & Era, 2006). Especially, center of pressure parameters described amplitude of sway such as mediolateral (ML) and anteroposterior (AP) displacement and velocities of them have been used for decades (Piirtola & Era, 2006). Recently, Shirado et al. (2004) found different COP distribution types between able-bodied and complete paraplegia patients. Namely, all able-bodied participants showed a central type pattern but all participants with SCI revealed a bilateral type. However, in more holistic perspectives about postural control, measuring of sway range with COP has some limitation in explanation of quantification of the regularity or predictability of a time series hidden in nonlinearity of human motion (Pincus, 1991). It was reported that dynamic structures in COP fluctuations in given tasks comprise the nonlinear properties in the postural system and it can be indicator of reorganization of degree of freedom, based on learning, physical changing, adaptation by injury, aging (Hong et al., 2008), pathology etc (Lindenberger et al., 2000; Marchese et al., 2003).

There are many secondary parameters derived from COP that have been reported as an outcome measures of postural control, such as mean sway amplitude, mean sway range from minimum to maximum, total excursion, root-mean-square (RMS) distance, mean sway

velocity, 95% ellipse sway area, etc (Baratto, Morasso, Re, & Spada, 2002; Duarte & Freitas, 2010; Raymakers, Samson, & Verhaar, 2005; Prieto et al., 1996).

2.4.2 Structural analysis

Various mathematical methods have been developed to analyze stabilographic data in under an assumption that COP migration is stochastic, chaotic or random walk. There are three main structural analyses techniques by perspectives on COP signal.

Collins and De Luca (1993) proposed that COP signal is a kind of stochastic processes modeled as random walk. They consider COP posturography as a random walk or Brownian motion controlled by two control system (a process of short term mechanism vs long term mechanism). The short term intervals in the variance of COP trajectory are changed by open-loop control schemes while the long term intervals in that of COP trajectory are utilized by closed-loop control schemes.

Baratto et al. (2002) proposed sway-density curve based on the idea that the postural stabilization is controlled by feedforward mechanism and therefore, the process of control is a sequence of anticipatory motor commands. Here, sway density curve means the counting the number of consecutive samples of COP trajectory that fall within a circle of known radius. Their idea is COP trajectories are not compatible with the Brownian movement.

2.4.3 Sample entropy

New perspectives of variability in chaos theory and nonlinear dynamics help us understand the problem of physiological processes of the neuromuscular system and dysfunction of brain (Van Emmerik, Rosenstein, McDermott, & Hamill, 2004). It is maintained that the observed amount of postural sway is not a direct indicator of postural stability and the larger sway is not related to a higher risk of falling. Researchers have begun to use nonlinear tools to examine postural control.

Recent studies have used several nonlinear tools simultaneously (Harbourne & Stergiou, 2009), based on the belief that this approach can verify the movement structure more clearly. Approximate Entropy (ApEn) is accepted as a useful method to quantify the regularity or dynamic structure in COP fluctuations (Pincus, 1991, 1995). ApEn measures the probability of reappearance of given patterns in time series. In this method, using time delay embedding, time series are reconstructed in a multi-dimensional state space with a given embedding dimension and a criterion of similarity.

To avoid bias in ApEn due to self-matches in the signal patterns, sample entropy (SampEn) was introduced . Vakorin et al. (2010) described the process to calculate the SampEn as follows.

> "Let x_t, $t = 1, ..., N$ where N is the number of data point, be realizations of a dynamical process μ_x.
>
> Next, we assume that the dynamics of the underlying m-dimensional system is reconstructed from the observed time series x_t using time delay embedding
>
> $$x_m(t) = (x_t, x_{t+1}, \cdots, x_{t-m+1})^{\mathrm{T}}$$
>
> for all $t = 1, N - m + 1$.
>
> Let $\theta(u)$ denote the Heaviside function, i.e. $\theta(u) = 1$ if $u > 0$, and $\|\cdot\|$ stand for the maximum norm distance between two delay vectors.
>
> The function
>
> $$B_i^m(x_m(i), r) = \frac{1}{N - m - 1} \sum_j \theta(r - \|x_m(i) - x_m(j)\|)$$
>
> for a given point i with the delay vector $x_m(i)$ reflect the number of points j such that the distance between the vectors $x_m(i)$ and $x_m(j)$ is less than r, excluding self-matches $i = j$. Similar to the $(m + 1)$-dimensional representation of x_t, the

function

$$A_i^m(x_{m+1}(i), r) = \frac{1}{N-m-1} \sum_j \theta(r - \|x_{m+1}(i) - x_{m+1}(j)\|)$$

for a given delay vector $x_{m+1}(i)$, is proportional to the number of points j such that the distance between the vectors $x_{m+1}(i)$ and $x_{m+1}(j)$ is less than r. Averaging the functions $B_i^m(x_m(i), r)$ and $A_i^m(x_{m+1}(i), r)$ across all the points $x_m(i)$ and $x_{m+1}(i)$, $i = 1, ...N-m$, respectively, we define

$$B^m(r) = \frac{1}{N-m} \sum_i B_i^m(x_m(i), r)$$

and

$$A^m(r) = \frac{1}{N-m} \sum_i A_i^{m+1}(x_{m+1}(i), r) \quad .$$

Then, the sample entropy is defined as

$$\text{SampEn}(m, r) = -\ln \frac{A^m(r)}{B^m(r)} \quad ."$$

Finally, SampEn can be interpreted in terms of the average natural logarithm of conditional probability that two delay vectors, which are close in m-dimensional space(meaning that the distance between them is less than the scale length r), will remain close in $(m+1)$-dimensional space. A greater likelihood of remaining close results in smaller values for the SampEn statistic, indicating less irregularities. Conversely, higher values are associated with the signals having more complexity and less regular patterns in their representations. Cross-sample entropy is a measure of "nonlinear temporal affiliation" between two signals. Conceptually, it assesses the dynamical properties of coupling between two time series.

Several studies have examined the effect of aging on the complexity of seated posture. For example, Harbourne and Stergiou (2003) showed that infants' adaptation process in

three different stages of sitting (entry, developing, and independent sitting). They argued that complexity was changed by stage it can be evidence of self-organizaing process of sitting. Also, the touch support manipulation has also been shown to reduce sitting postural sway in infants learning to walk independently (Chen, Metcalfe, Jeka, Clark, & E., 2007). This suggests that even at a fairly early stage in motor development, augmented sensory information has the potential to reduce postural sway. Hong et al. (2008) found that the COP path length and complexity decreased with age. They argued that the postural sway of the younger children showed slower fluctuations that were more tightly coupled across the axes of motion when compared to that of adults.

Contrary to the evidences of a general decrease in complexity with aging, Vaillancourt and Newell (2002); Vaillancourt et al. (2004) argued that complexity should be changeable by not only aging, but also the task. The researchers showed that for a task where participants maintained a constant level of isometric force, aided by visual feedback, complexity decreased with aging (Vaillancourt et al., 2004). However in the tracking a sinusoidal wave of same participants by varying isometric force, the complexity increased with aging.

Even though there have been studies on several populations for sitting postural control, it is not known if functional limitations from spinal cord injury affect the complexity in postural control in nonlinear dynamic approach. This investigation seeks to fill in this knowledge gap by examining the differences in seated postural control among spinal cord injury patients of two different level and non spinal cord injury patients.

2.4.4 Virtual time contact to functional boundary

An ecological approach to determine the virtual time to contact (VTC) to the stability boundary for the postural stability estimation of human standing posture were introduced by Slobounov, Slobounova, and Newell (1997).

For the VTC calculation, a position vector of the COP on a virtual trajectory $\tau_i(t)$ was determined for each moment in time t_i based on the instantaneous COP velocity and

acceleration:

$$x, y(\tau) = r_{x_i,y_i}(t_i) + v_{x_i,y_i}(t_i)\tau + a_{x_i,y_i}(t_i)\tau^2/2$$

where $r_{x_i,y_i}(t_i)$ is the instantaneous position vector, $v_{x_i,y_i}(t_i)$ is the instantaneous velocity vector and $a_{x_i,y_i}(t_i)$ is the instantaneous acceleration vector in the x and y directions.

With the current virtual trajectory, the position vector for crossing point (x_c, y_c) were determined by

$$\left(\frac{x_c}{R_x}\right)^2 + \left(\frac{y_c}{R_y}\right)^2 = 1$$

where R_x and R_y are the semiaxes of the ellipse traced out by the initial pivoting.

In this approach, researchers measure functional boundary by "without having to initiate a step, the leaning forward, pivoting at the ankle joint, and proceeding in a circular direction leaning as far as possible" (Haibach, Slobounov, Slobounova, & Newell, 2007, p. 473). Their consecutive studies verified that VTC is more sensitive tool than traditional measures in that it shows higher coefficient of variation than those of velocity and acceleration and less time to contact with advanced age in aging spectrum.

Chapter 3

Complexity in seated postural control in individuals with spinal cord injury

Abstract

The complex output of the neuromuscular system results from the interaction of multiple control processes operating at unique timescales. The loss of complexity hypothesis maintains that if a physiological system (e.g. neuromuscular) is damaged (i.e. decrease in control processes), the output would be less complex. However, there is growing evidence that physiological complexity does not simply decrease with damage, but rather, changes in complexity are driven in part by task constraints. This proposition forms the tenets of the loss of adaptability hypothesis: that the observed complexity of a system depends on the tasks being performed and that healthy function is characterized by the adaptability of the output to task constraints.

Given that seated postural control is not a simple passive alignment of body segments, but rather a complex motor task dependent on sensorimotor integration and individuals with SCI are an ideal population to test the predictions of the loss of adaptability hypothesis because they have a defined biomechanical and neural DoF, seated postural control in persons with SCI would be a prime motor task to test the predictions of loss of adaptability hypothesis. This study examined how the complexity of seated postural control is influenced by SCI across different task constraints.

Methods

Twenty participants (10 SCI and 10 controls) participated in this study. To quantify seated postural control, participants sat on a wooden box placed on a force platform (AMTI, Inc.) with their arms by their side. The center of pressure (COP), a reflection of the neuromuscular response to the imbalances of the body's center of gravity was quantified along the anterioposterior (AP) and mediolateral (ML) axes. Participants completed three different sitting tasks (i.e. sitting still, self-paced rocking and fixed rhythmic rocking). Sway range and a nonlinear technique (Sample entropy) in both AP and ML direction were measured to quantify variability and complexity of COP time series respectively.

Results

Overall individuals with SCI had larger variability and less complexity in their COP dynamics in sitting still than controls. However, there were no differences of complexity in their postural control in self-paced rocking and fixed rhythmic rocking (rocking at 50 beat/min) tasks between persons with and without SCI.

Conclusion

The observation that the complexity of seated postural control in persons with SCI is dependent on the intrinsic dynamics of the task is congruent with the loss of adaptability hypothesis. The current observations highlight the limitations of the loss of complexity theory in neuromuscular output which was theorized that the complexity of physiological output is decreased with age or disease.

3.1 Introduction

3.1.1 Physiological Complexity

Physiological function results from a complex interaction of multiple control mechanisms operating at various levels (cellular, organ-level and systematic) across unique time scales. The continuous dynamic interaction of control processes allows for physiological adaptability to internal and external stressors (Lipsitz & Goldberger, 1992). The dynamic structure of physiological output is not detected by standard distributional statistics. Rather, nonlinear analysis is needed to quantify the time evolving properties of physiological output (Stergiou, 2004). To demonstrate the difference between distributional statistics and nonlinear statistics, heart rate (HR) time series of a young and elderly adult plotted in Figure 3.1. Although it is visually clear in Figure 3.1 that there are differences between heart rate traces of the young and old subjects, traditional distributional metrics are nearly identical ($HR_{young} = 64.7 \pm 3.9\,bpm$; $HR_{old} = 64.5 \pm 3.8\,bpm$). However, a clear difference between heart rate traces is observed when a measure capable of quantifying the complexity of the times series, such as approximate entropy (ApEn) is utilized ($ApEn_{young} = 1.09$; $ApEn_{old} = 0.48$) (Lipsitz & Goldberger, 1992). ApEn ranges from zero to two with relatively higher ApEn value indicating that physiological signal is more complex. In addition to highlighting the importance of quantifying complexity of a signal, Figure 3.1 also illustrates a loss of complexity (i.e. dynamic structure) that presumedly accompanies aging.

Lipsitz and Goldberger (1992) propose that, with advanced age and disease, there is a decrease in physiological complexity that results in functional impairment. Based on data from cardiovascular and respiratory systems, they maintain that, as complexity decreases, the output signal becomes more regular and less complex, coinciding with functional decline.

There are several examples of loss of complexity with advanced age. For instance, examples of loss of complexity related to aging and disease include reduced heart rate variability, a reduction in nerve conduction velocity, a loss of high frequency sound, and a reduction of

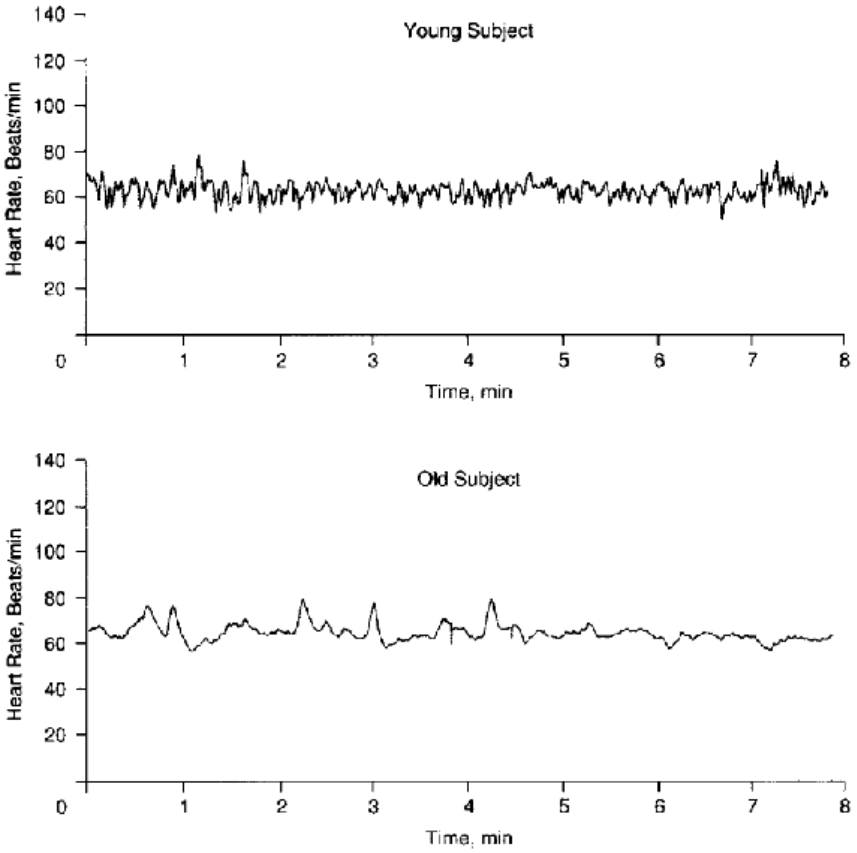

Figure 3.1: Heart rate time series for a young female (22 years old) (mean: 64.7; SD: 3.9) and an old male (73 years old) (mean: 64.5, SD: 3.8); adapted from Lipsitz and Goldberger (1992, p. 1807).

irregularity in postural sway (Kaplan et al., 1991; Mader, 1984; Manor et al., 2010; Munsat, 1984). Most cases of loss of complexity in the human physiological system are the results of the interaction of multiple physiological control systems. For instance, heart rate variability may be caused by reduced sinus node cells, altered β-adrenoceptor responsiveness, or reduction in parasympathetic tone (Lipsitz & Goldberger, 1992).

To measure changes in physiological complexity, one of the most well-known methods is to measure entropy of a system. Originally, the term, "entropy" was used in the field of thermodynamics, which is the science that deals with formation, transformation of, and interaction of heat. In thermodynamics, the entropy in the micro level is essentially a statistical approach in which probability theory and statistical models are used to determine

most probable or average values of the velocity and position of the elemental particles of the atom or molecule. Pincus (1991) developed the approximate entropy algorithm to measure regularity or predictability of the system. It is considered a useful tool to measure level of loss or impairment of functional components.

3.1.2 Loss of complexity vs. loss of adaptability

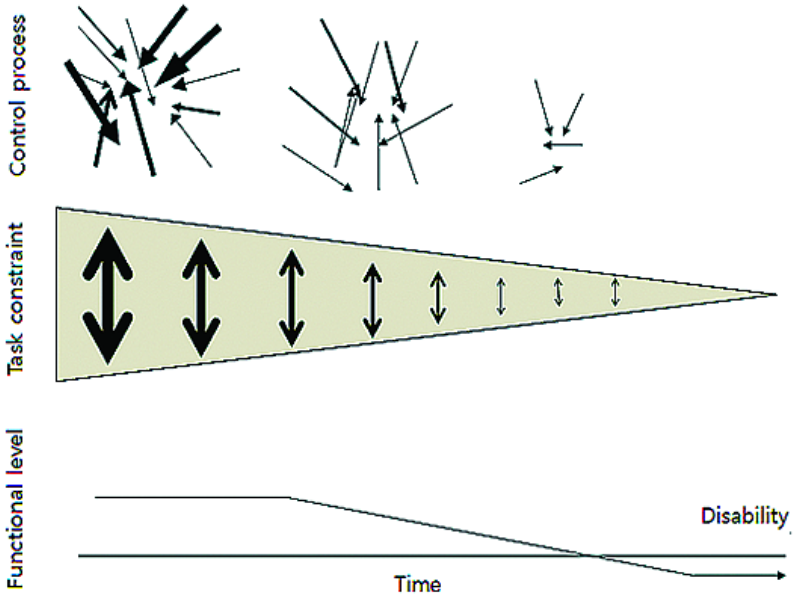

Figure 3.2: A schematic of loss of adaptability hypothesis. Multiple interacting control process (top) was decided by functional level (bottom) and task constraints (middle) in which function may fall to the critical level below which an organism can no longer adequately respond to stress; modified from Lipsitz (2002).

Vaillancourt and Newell (2002); Vaillancourt et al. (2004) proposed that the "loss of complexity" hypothesis is too narrowly generalized to slow timescales of aging and disease progression. They highlighted that when task dynamics are considered, in many cases, physiological output does not always demonstrate a loss of complexity with age, but rather an inability to "match" output to the task constraints Figure 3.2. Figure 3.2 depicts the loss of adaptability hypothesis. The top panel represents the number of control processes interacting to produce motor output. The middle panel indicates that the observed complexity (i.e.

triangle) is not static but rather determined by the task constraints (i.e. arrows). The bottom panel depicts that the amount of control process is functionally relevant, and that, with a decrease in adaptability, an individual demonstrates a loss of function.

They refer to this inability to constrain output to task dynamics as a loss of adaptability. The "loss of adaptability" theory maintains that, the physiological functional change, due to aging or impairment reduces the capacity of the system to adapt to various forms of stressors that are defined by time-varying task constraints (Sosnoff & Newell, 2008).

Some of the first evidence in support of loss of adaptability come from Vaillancourt and Newell (2003); Vaillancourt et al. (2004). They demonstrated that there can be bidirectional changes in complexity according to the confluence of constraints in action (i.e. whether a task is "fixed point dynamic" (i.e. constant force task) or "oscillating intrinsic dynamic" (i.e. sinusoidal force task)). That is, when subjects were required to maintain a constant isometric force, the structure of force output in older-old adults (i.e. over 75-year-old-persons) was less complex. However, when older-old adults were required to produce force rhythmically, the older adults had more complex force pattern compared to young adults (Vaillancourt & Newell, 2003). Similarly, most studies of aging and disease related changes in neuromuscular output have emphasized "loss of complexity" phenomenon using fixed point dynamics (Lipsitz & Goldberger, 1992).

3.1.3 Complexity of COP time series

As a physiological output, oscillations in the center of pressure (COP) time series, a reflection of the system's neuromuscular response to the imbalances of the body's center of gravity (Winter, 2005), are indicative of physiological rhythms resulting from the integration of multi-sensory information and motor output (Cavanaugh, Guskiewicz, & Stergiou, 2005; Oie, Kiemel, & Jeka, 2002; Newell, 1998; Hong et al., 2006). As such, postural control lends itself to complexity analysis. Examinations of COP profiles have revealed that special populations have distinct variability and complexity characteristics. For example, in sitting

postural control, there were reduced complexity in COP dynamics among people with mental retardation (Berkson & Andriacchi, 2000; Hong et al., 2006), and development (Hong et al., 2008; James et al., 2009), and pathology (i.e. Parkinson's disease) (Schmit et al., 2006). Therefore, measuring COP dynamics during sitting is an accepted experimental paradigm to examine the loss of adaptability hypothesis (Hong et al., 2006, 2008).

3.1.4 Tasks in sitting dynamics

Due to the importance of sitting, several investigations have examined the effects of aging and disease on the complexity of seated posture. These investigations utilize two distinct sitting tasks: sitting still, and body rocking (Hong et al., 2006, 2008; James et al., 2009). It is maintained that sitting still is characterized by a fixed point dynamic, while body rocking is characterized by intrinsic rhythmical dynamic. Even though these tasks have different intrinsic dynamics, they can be quantified by measuring the COP (Latash, Ferreira, Wieczorek, & Duarte, 2003). An experimental example of the exploitation of the distinct intrinsic dynamics is in Hong et al. (2006), who report that persons with stereotypic movement disorder (i.e. mental retardation) had less complexity in sitting still (i.e. lower approximate entropy value) compared to healthy controls, while there were no differences in body rocking tasks. These results imply complexity of postural control depends on task dynamics and unidirectional loss of complexity perspective should be reconsidered with tasks having different intrinsic dynamics (i.e. fixed point intrinsic vs. oscillating intrinsic dynamics). Their findings are preliminary evidence to support the loss of adaptability view to aging, disease and functional decline.

3.1.5 Spinal cord injury and complexity in sitting

Human movement results from the dynamic interplay of a large number of degrees of freedom – which afford complexity and adaptability of action (Latash, 2008). Even though most of

research testing the loss of complexity theory has assumed differences in DoF between populations (Vaillancourt & Newell, 2002; Mayer-Kress, Liu, & Newell, 2006), the actual number of DoF in a given population is rarely defined. For instance, although older adults' reduced complexity in constant muscular force output has been suggested to result from a decline in active DoF, there is no direct evidence that older adults have a decreased DoF (Lipsitz & Goldberger, 1992; Newell, 1998). Individuals with SCI are an ideal population because persons with SCI have a defined biomechanical and neural DoF determined by injury level.

This investigation was designed to understand how the complexity of sitting postural control in individuals with SCI is affected by task constraints. To test the predictions of the loss of adaptability hypothesis, persons with and without SCI performed several sitting postural control tasks (quiet sitting, self-paced body rocking, and fixed-paced body rocking). The quiet sitting task is conceptualized as having a fixed point intrinsic dynamic, while the body rocking tasks are seen as having an intrinsic rhythmic dynamic. It was hypothesized that persons with SCI would demonstrate less complexity than persons without SCI during quiet sitting, whereas persons with SCI would demonstrate more complexity than persons without SCI during their fixed-paced body rocking.

3.2 Methods

3.2.1 Participants

A total of 20 persons, 10 individuals with spinal cord injury (7 males and 3 females) and 10 able bodied age, gender, sitting height matched controls took part in this study. Persons with SCI were limited to paraplegia. Injury levels of the participants ranged from T4 to L4. The injury level of the 10 individuals with SCI was "complete" spinal injury between L5 to T7 which refers to "A" level by American Spinal Injury Association. Experimental procedures were outlined to the subject prior to the beginning of the experimental session, and all participants provided written consent prior to the start of testing. IRB consent

procedures were outlined to the subject prior to the beginning of the experimental session.

3.2.2 Procedures

All participants underwent seated postural control assessment. Seated postural control assessment consisted of three separate tasks: quiet sitting, self-paced body rocking and fixed rhythmic rocking. Quiet sitting consisted of sitting as still as possible for 30s. Self-paced rocking consisted of participants performing at their self-selected paced rhythmically rocking their upper body in anteroposterior (AP) direction for 30s. Fixed rhythmic rocking consisted of participants rocking their upper body along the AP axes at 50 beat/min pace (0.83 Hz) (Latash et al., 2003). An auditory metronome signal was used to provide the rhythm. All sitting tasks were carried out with participants sitting on an AMTI OR6-3A (American Mechanical Technology, Inc., Watertown, MA) force platform on top of a custom built wooden box. Each unique condition was performed twice and used average values.

3.2.3 Data analysis

The force platform records six components of postural dynamics including three force components: mediolateral force (Fx), anteroposterior force (Fy), and vertical force (Fz); and three moment components taken about the about the respective axes: M_x, M_y, M_z. The signals were amplified using a six-channel AMTI-Model MSA6 Strain Gage Amplifier. A gain of 2000 was used. The bridge excitation was set to 10 V and data were collected at 100 Hz.

Signals from the force plate were filtered with a 4th order low pass Butterworth filter with an adequate cut-off frequency. The adequate cut-off frequency was determined with residual plot analysis (Winter, 2005). The COP was separately calculated along with anteroposterior

(AP) and mediolateral (ML) axes by using following equations:

$$COP_{AP} = (-h \times F_x - M_y)/F_z$$

$$COP_{ML} = (-h \times F_y + M_x)/F_z$$

where h is the offset between the force plate sensors and the surface ($h = 20.6\,\text{mm}$).

Variability of the COP time series in both AP and ML directions was calculated utilizing sway range (Prieto et al., 1996). Sway range was defined as the absolute value of the difference between the maximal and minimal values for the given direction:

$$\text{Sway Range} = |\text{COP}_{\max} - \text{COP}_{\min}|$$

Complexity of the COP time series in both AP and ML directions was calculated utilizing sample entropy. This method was developed by Richman and Moorman (2000). Sample entropy (SampEn) of COP time series both in AP and ML direction were calculated by the negative natural logarithm of the conditional probability that repeated patterns of length m are similar to $m + 1$:

$$\text{SampEn}(m, r, N) = -\ln \frac{C_{m+1}(r)}{C_m(r)}$$

where C is the probability that two points within m, the window of comparison, are similar. Similarity is defined as the points having a difference of less than r, the radius of similarity. r was determined by 15% of the SD of COP time series (Costa, Goldberger, & Peng, 2003). A low value of the SampEn indicates that the time series is more deterministic, predictable, or less complex, and a high value indicates randomness, less predictable, or more complex.

3.2.4 Statistical analysis

Each dependent variable (i.e. sway range and SampEn) was placed into a two-way (2×3) mixed model of ANOVA with *group* (Control and SCI group) as the between subject factor

and task (Quiet sitting, self-paced rocking and fixed-rhythm rocking) as the within subject factor. A separate independent t-test for each variable was conducted to examine interaction effects between group and task. While the α-level for significance was ≤ 0.05, a Holm-modified Bonferroni correction was applied to control for type-I error caused by multiple comparisons. Effect sizes associated with F-ratios were expressed as partial eta squared (η_p^2). All data were analyzed using SPSS version 18.0 (SPSS Inc., Chicago, IL., USA).

3.3 Results

3.3.1 Sway Range (AP)

Statistical analysis on sway range in AP direction revealed that there was a significant main effect of task, but not group (Table 3.1). Overall, quiet sitting (5.37 ± 0.82 mm) had a smaller mean sway range than self-paced rocking (83.31 ± 7.64 mm) and fixed-rhythmic rocking (79.38 ± 7.91 mm) ($p < 0.01$ in both comparisons). However, there were no significant difference between self-paced rocking and fixed-rhythmic rocking ($p = 0.72$). There was no significant interaction between group and task (Table 3.1).

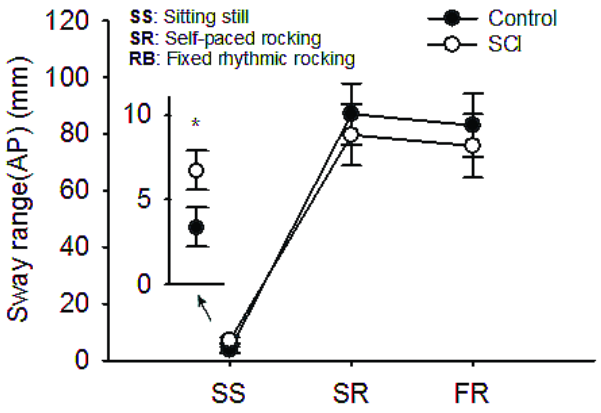

Figure 3.3: Mean sway range (AP) of COP as a function of group during sitting still, self-paced rocking and fixed rhythmic rocking. Error bars represent standard error of the mean.

Independent t-tests showed that SCI group had larger mean sway range than control in

the quiet sitting (7.04 ± 1.55 vs. 3.70 ± 0.51; $t(1, 18) = -2.03$, $p = 0.05$). However there were no differences in self-paced rocking (79.60 ± 12.86 (SCI group) vs. 87.03 ± 8.24 (controls); $t(1, 18) = 0.48$, $p = 0.63$) and fixed-rhythmic rocking (83.00 ± 7.64 (SCI group) vs. 75.76 (controls) ± 13.84; $t(1, 18) = 0.45$, $p = 0.65$) (Figure 3.3).

3.3.2 Sway Range (ML)

Statistical analysis on sway range along the ML axis revealed that there was a significant main effect for group and task (Table 3.1) . Overall, subjects with SCI had larger mean sway range (ML) than controls (22.51 ± 3.77 vs. 7.91 ± 3.77 mm). Overall, quiet sitting (4.43 ± 0.74 mm) had a smaller mean sway range than self-paced rocking (20.10 ± 3.68 mm) and fixed-rhythmic rocking (21.09 ± 4.00 mm) ($p < 0.01$ in both comparisons). However, there were no significant difference between self-paced rocking and fixed-rhythmic rocking (p=0.96).

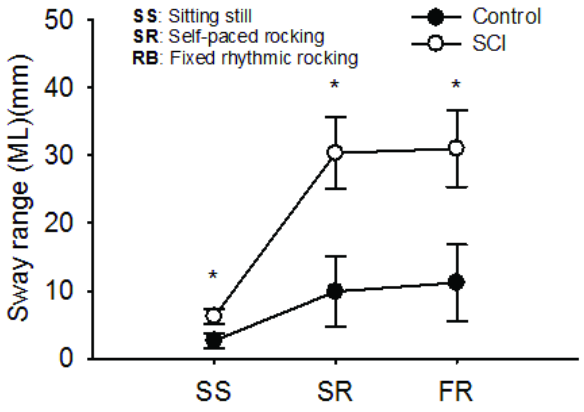

Figure 3.4: Mean sway range (ML) of COP as a function of group during sitting still, self-paced rocking and fixed rhythmic rocking. Error bars represent standard error of the mean.

There was an interaction between group and task (Table 3.1). Independent t-tests showed that SCI group had larger mean sway range (ML) than control in the quiet sitting (6.22 ± 1.40 vs. 2.65 ± 0.47; $t(1, 18) = -2.41$, $p = 0.02$), self-paced rocking (30.33 ± 0.85 vs. 9.88 ± 0.85; $t(1, 18) = -2.77$, $p = 0.01$) and fixed-rhythmic rocking (30.99 ± 7.93 vs. 11.20 ± 1.04;

$t(1, 18) = -2.47$, $p = 0.02$) (Figure 3.4).

Table 3.1: Summary of analysis of variance for variability and complexity measures.

Independent variable	Range_{AP}			Range_{ML}			$\text{SampEn}_{\text{AP}}$			$\text{SampEn}_{\text{ML}}$		
Main effect	F-ratio	p-value	η_p^2	F-ratio	p-value	η_p^2	F-ratio	p-value	η_p^2	F-ratio	p-value	η_p^2
Group	0.13	0.71	< 0.01	7.46	**0.01**	0.29	5.48	**0.03**	0.22	9.03	$<$ **0.01**	0.33
Task	89.97	$<$ **0.01**	0.83	19.98	$<$ **0.01**	0.52	14.75	$<$ **0.01**	0.45	30.22	$<$ **0.01**	0.62
Interaction												
Group × task	0.44	0.64	0.02	5.22	**0.01**	0.22	**6.13**	$<$ **0.01**	0.25	0.57	0.56	0.03

η_p^2 is partial eta squared.

3.3.3 SampEn (AP)

Statistical analysis on SampEn in the AP direction revealed that there was a significant main effect for group and task (Table 3.1). Overall, SCI group had smaller SampEn along the AP direction than controls (0.20 ± 0.01 vs. 0.26 ± 0.01). Also, overall, quiet sitting had a larger mean SampEn than self-paced rocking and fixed-rhythmic rocking (0.31 ± 0.02 vs. 0.18 ± 0.01, vs. 0.20 ± 0.01, respectively). There was an interaction between group and task (Table 3.1 and Figure 3.5). Independent t-tests showed that SCI group had smaller SampEn than control (0.42 ± 0.04 vs. 0.27 ± 0.02; $t(1, 18) = 2.74$, $p = 0.01$) in the quiet sitting. However, there were no differences in self-paced rocking ($t(1, 18) = 0.80$, $p = 0.43$) and fixed rhythmic rocking ($t(1, 18) = -0.71$, $p = 0.48$) between two groups (Figure 3.5).

3.3.4 SampEn (ML)

Statistical analysis on SampEn along the ML axis revealed that there were significant main effects for group and task (Table 3.1). However, there was no significant interaction between group and task (Table 3.1 and Figure 3.6). Overall, subjects with SCI had smaller mean SampEn than controls (0.25 ± 0.03 vs. 0.38 ± 0.03). The mean SampEn in self-paced rocking (0.35 ± 0.02) and fixed rhythmic rocking (0.36 ± 0.02) were larger than that in sitting still (0.24 ± 0.02) ($p < 0.01$ in both comparisons). There were no differences between self-paced rocking and fixed rhythmic rocking ($p = 0.97$).

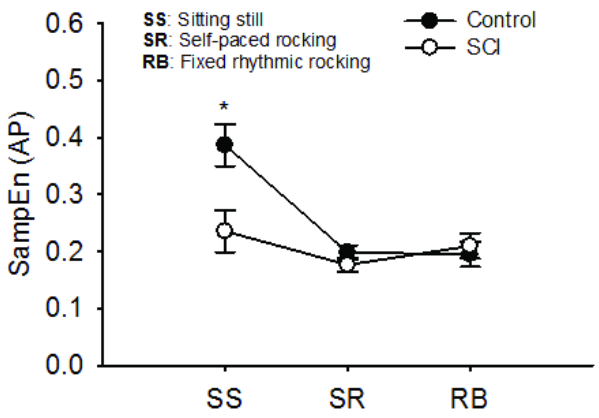

Figure 3.5: Mean SampEn of COP along AP axis during sitting still, self-paced rocking and fixed rhythmic rocking for control group and SCI group. Error bars represent standard error of mean.

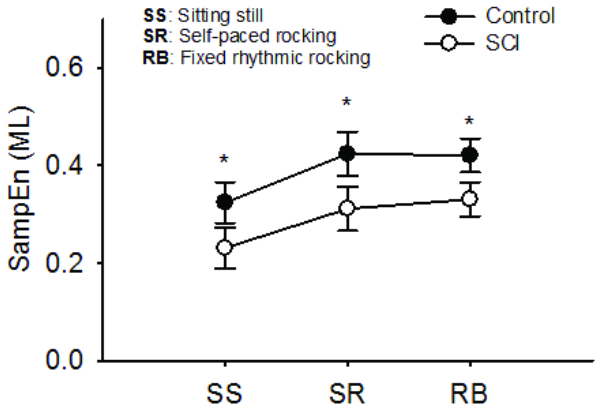

Figure 3.6: Mean SampEn of COP along the ML axis during sitting still, self-paced rocking, and fixed rhythmic rocking for control group and SCI group. Error bars represent standard error of mean.

3.4 Discussion

The current investigation examined the predictions of the loss of adaptability hypothesis utilizing a series of seated postural control tasks in persons with SCI and controls. The loss of adaptability hypothesis maintains that, with declines in function, physiological complexity is determined by the ability to respond to environmental stressors (e.g. task constraints). It is also maintained that functional decline is characterized in part by lower performance in a given task.

Overall, persons with SCI had larger variability in the sitting still condition indicated by larger sway range in both AP and ML axis than control participants. However, there were no significant group differences in variability in self-paced rocking and fixed rhythmic rocking in AP direction. Those results support the idea that the magnitude of variability is influenced by task constraints.

The reduced seated postural control performance in the SCI group did not coincide with decreased complexity. Rather the SCI groups seated postural control complexity was dependent on task constraints. It was observed that persons with SCI showed less complex postural dynamics (i.e., SampEn) in the sitting still in the AP direction than persons without SCI. There were no group differences in postural control complexity along the AP axis in the body rocking conditions. In contrast, the SCI group also had lower complexity than the control group along the ML axis. Moreover, there was no effect of task constraints on the complexity of movement along the ML axis. In other words, the interaction between group and task only occurred along the axis in which the body rocking was conducted. These observations have important implications for the loss of adaptability hypothesis.

The current findings of reduced postural sway complexity in quiet sitting in persons with impairment (e.g. SCI) compared to non-impaired control participants are congruent with previous reports focusing on persons with impaired seated postural control (Hong et al., 2006; Perlmutter et al., 2010). Importantly, these previous investigations focused on population

groups with cortical impairment (chronic stroke and mental retardation). The observation of the current investigation extends this previous research in demonstrating that this loss of complexity in quiet sitting is not purely due to cortical impairment but apparently also when there is any neuromuscular dysfunction.

Also importantly, although there were group differences between the SCI group and control group in complexity of quiet sitting, there were no differences in the body rocking tasks along the AP axis. The effect of task constraints is consistent with the report on Hong et al. (2006). They demonstrated that there were no differences in complexity in body rocking between persons with stereotyped movement disorder and mental retardation and controls, whereas controls had larger complexity in sitting still. They argued that body rocking is an inherently easier motor control task requiring less control.

In previous examinations of the loss of adaptability hypothesis, differences between groups neurological function was simply assumed and not quantified. For instance, Sosnoff and Newell (2008) showed that older adults had lower complexity output in a constant isometric force task, while they had higher complexity output in rhythmical force tasks. It was assumed that older adults had fewer neurophysiological DoF than the younger adults, but no quantification of this assumption was performed. By examining motor output of persons with SCI, this investigation overcomes this implicit assumption and insures that the comparison between groups was a true comparison between groups with different level of biomechanical and neurological degrees of freedom.

The body rocking tasks, both the self-paced and fixed-rhythmic rocking conditions, required participants to modulate their posture only in the AP direction. This is an important experimental detail, given that the effect of task constraints was only observed on complexity in the AP axis and not the ML axis. In other words there was an interaction between group and task constraints only in the direction specified by task instructions. Not only is this observation congruent with predictions of the loss of adaptability hypothesis, but it also indicates that participants are capable of constraining their seated postural control along a

given axis while ignoring sway in the orthogonal direction. This also raises the possibility that other theoretical approaches concerning human movement could be applied to seated postural control. For instance, the uncontrolled manifold approach could be applied in future work to examined seated postural control in persons with SCI (Latash, 2008). The uncontrolled manifold approach maintains that variability of movement is only controlled in a task defined manner and movement variability in non-task define axis is allowed to fluctuate freely.

Although the theoretical import of the current investigation is clear, it is possible that the observations also have clinical significance. Lower amounts of complexity in sitting still indicates that persons with SCI have a reduced motor repertoire in which to select actions to accomplish the task (Morrison, Colberg, Parson, & Vinik, 2010). A potential result of this reduced motor repertoire is an inability to adapt their neuromotor output to environmental stressors (e.g. task constraints). Consequently, the reduced postural control adaptability could have real world implications. For instance, Morrison et al. (2010) demonstrated that persons with Diabetes who have a history of falling have reduced standing postural control complexity. It is possible that the reduced adaptability of seated postural control is a risk factor for falls in persons with SCI. This possibility warrants further investigation.

3.5 Limitation of study

Despite the novel findings of this investigation there were several empirical limitations. First, most of the participants in the SCI group were wheelchair athletes. So the generalization of the current findings to the general SCI population is suspect. In an effort to minimize fatigue the order of the seated postural conditions was not randomized. It is possible that participants were more fatigued in the body rocking conditions than the sitting still conditions. Future work should randomization conditions to minimize this possibility. The lack of randomization also could have influenced the learning of the distinct motor tasks. The lack

of difference between the body rocking and paced body rocking condition could potentially result from this experimental limitation.

3.6 Conclusion

Even though there has been a significant amount of support for the loss of complexity theory, there has been increasing evidence that this theory is an oversimplification. The present study demonstrates that persons with SCI had less complexity of seated postural control only in a task characterized by a fixed point dynamic (e.g. sitting still). However, in a rhythmical task characterized by an oscillating intrinsic dynamic such as body rocking there were no group differences in complexity. These observations are congruent with the loss of adaptability hypothesis. Further work determining the clinical importance of these observations is warranted.

Chapter 4

Effect of fingertip contact in postural control of individuals with spinal cord injury

Abstract

The complexity of postural control results from a dynamic interplay of multiple sensory inputs, including visual, vestibular, proprioceptive and appropriate muscular action. It is well known that a light fingertip touch on a stationary surface enhances standing postural control in various populations. This beneficial effect is believed to stem from utilization of additional sensory information. The effect of additional sensory information on seated postural control in persons with or a without SCI is not clear. It is logical to assume that there would be a greater influence of additional sensory information on the seated postural control (i.e. reduced sway, increased complexity) in individuals with SCI compared to individuals without SCI. The purpose of this experiment is to determine if finger contact influences variability and complexity of seated postural control in quiet sitting of individuals with SCI.

Method

Sixteen subjects (8 SCI and 8 controls) participated in this study with the same experimental as described in Chapter 3. Sitting balance was assessed in three conditions of fingertip contact were tested (i.e. no touch, light touch (less than 1N), and heavy touch (over 1N touch). Participants were tested in two different sitting surface conditions (stable condition and unstable condition). Traditional measures and complexity measures of COP were calculated.

Results

The additional cutaneous sensory information had a minimal effect on postural complexity of persons with SCI. However, there was a reduction in postural sway with additive sensory information in the SCI group. The beneficial effect was greater in the unstable surface condition.

Conclusion

Additional cutaneous sensory information reduces the variability, but not complexity of the sitting balance in persons with SCI. This observation highlights the difference in variability and complexity metrics. This finding indicates that additional cutaneous information is beneficial in persons not only in standing postural control but also in seated postural control. The ecological benefit of this observation remains to be determined.

4.1 Introduction

4.1.1 Sensory information and postural control

Postural control results from a dynamic interplay of multiple sensory inputs, including visual, vestibular, and proprioceptive inputs and appropriate muscular action (see Figure 4.1) (Jeka et al., 1996; Massion, 1984). Importantly, the neuromotor system can change its dependence upon a given sensory input in order to maintain a posture (Horak & Macpherson, 1996; Nashner, Black, & Wall, 1982). For instance, one investigation demonstrated that with the addition of light fingertip contact to a stable surface, there was minimal differences in upright postural control between individuals with vestibular dysfunction and controls (Creath, Kiemel, Horak, & Jeka, 2002).

Effects of sensory information on complexity and variability of postural control has been studied utilizing platform posturography in which force platforms measure fluctuations in

the ground reaction force point of application (i.e. center of pressure (COP)) (Winter, 2005). It has been repeatedly observed that postural control is enhanced (i.e. more complex or less variable) when more sensory information is available compared to less sensory information (Bolbecker et al., 2011; Creath et al., 2002; Horak, 2009; Manor et al., 2010) (See Figure 4.1).

4.1.2 Finger tip contact and compensation in postural control

Research has demonstrated that among somatosensory inputs, haptic inputs from cutaneous and kinesthetic receptors are important for postural control (Baccini et al., 2007; Jeka et al., 1996). There are several investigations concerning the role of cutaneous inputs on postural control. A common manipulation of cutaneous input is to provide light fingertip contact to a stationary object during a postural control task. Light finger tip contact with a stationary surface enhances standing postural control in various populations (e.g., young and old adults, individuals with anterior cruciate ligament injury (ACL), Diabetes Mellitus, blindness, bilateral vestibular loss) (Baccini et al., 2007; Backlund-Wasling, Norrsell, Göthner, & Olausson, 2005; Bonfim, Grossi, Paccola, & Barela, 2008; Dickstein, Shupert, & Horak, 2001; Jeka & Lackner, 1994; Jeka et al., 1996; Kouzaki & Masani, 2008; Lackner, Rabin, & DiZio, 2000).

This beneficial effect of additional cutaneous information tends to be greater in persons with more impaired postural control. For example, Baccini et al. (2007) reported that the effectiveness of light touching in reducing postural sway was greater in older adults than young adults. Dickstein et al. (2001) also reported that the effectiveness of light touch in decreasing sway was larger in patients with Diabetes Mellitus than in age-matched controls. In both investigations, it was argued that haptic cues from finger touch counterbalanced age or disease related sensory loss. Despite the beneficial effect of additional cutaneous information on postural control in various clinical populations, there are no extant data on the effect of additional cutaneous information on seated postural control in persons with

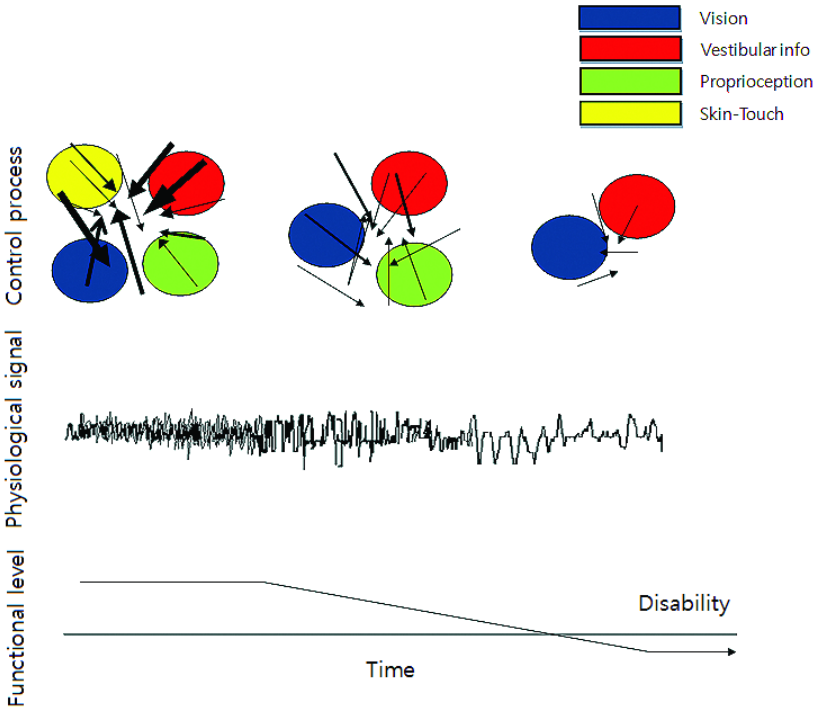

Figure 4.1: Schema of additional sensory information, and complexity and variability; modified from Lipsitz (2002).

SCI.

The research on seated postural control in SCI has focused mainly on motor output necessary to maintain a seated posture. For instance, it has been observed that individuals with SCI compensate for their loss of postural muscle activity (due to injury) by using non-postural muscles such as latissimus dorsi and trapezius pars ascendens to control their upper body in sitting (Seelen et al., 2001). Although it is known that there is decreased seated postural control (e.g. greater postural sway) in persons with SCI, their ability to utilize sensory information during a seated postural control task is not well understood.

Previous research on seated postural control in healthy adults has examined the effect of unstable support surface (Lanzetta, Cattaneo, Pellegatta, & Cardini, 2004; Reeves, Everding, Cholewicki, & Morrisette, 2006; Silfies, Cholewicki, & Radebold, 2003). It is maintained that sitting postural control on an unstable support surface is a more challenging postural task

than maintaining a seated posture on a firm stable surface (Lanzetta et al., 2004). It was assumed that the beneficial effect of additional sensory information would be paramount in a challenging postural control task.

The main theoretical concept being tested in this chapter is depicted above (Figure 4.1). To quantify the effect of additional sensory information on the physiological signal, traditional measures and complexity measures of COP were calculated. The figure illustrates how sensory information corresponds with variability, complexity and functional ability. The left aspect represents the availability of sensory information with the greatest amount of inputs coinciding with the smallest amount of variability, greatest complexity and functionality. In contrast the fewest sensory inputs shown on the right aspect of the figure corresponds with the greatest amount of variability, lowest complexity and lowest function. The middle panel suggests that if additional cutaneous sense (skin-touch) is provided as depicted upper panel left figure, the postural complexity would increase and variability would decrease.

With this framework in mind, the purpose of this current investigation was to examine the effect of additional cutaneous sensory information on seated postural control in persons with and without SCI in stable and unstable conditions. It was hypothesized that persons with SCI would have lower complexity and greater variability in their seated postural output (i.e. COP) compared to controls when no additional sensory information was available. It was also predicted that persons with SCI would have an increase in complexity and a decrease in variability of postural output with additional sensory information and that the effect on the participants' postural control would be minimal. Lastly, it was hypothesized that the benefit of the additional sensory information would be the greatest when the sitting surface was unstable.

4.2 Methods

4.2.1 Participants

Sixteen young adults (age range 19-35 yr) participated in this study. They were 8 spinal cord injured individuals (lesion between T4 and L4) and 8 able bodied age, gender, sitting height matched controls. The able-bodied controls had no known musculoskeletal injuries or neurological disorders that may affect their seated postural control. Persons with SCI also had no other injuries which would influence seated postural control. Written informed consent was obtained from all subjects prior to testing. IRB consent procedures were outlined to the subject prior to the beginning of the experimental session.

4.2.2 Procedures

In order to record seated postural control, participants sat on the force plate placed on custom built wooden box. The box had no back support and sufficient elevation so that the participants' feet did not touch the ground. Participants were required to stay still until the researcher gave a verbal signal that the trial was completed.

In order to access the effect of additional sensory information on seated postural control, three levels of applied force from fingertip contact were used (Baccini et al., 2007).

(i) No touch (NT): participants sat with both arms resting on the hip.

(ii) Light touch (LT): participants were limited to a maximum of 1N of applied force through right index finger.

(iii) Heavy touch (FT): participants were directed to exert as much force as necessary through their right index finger on a stable surface while sitting still.

When participants began light touch and force trials, they were asked to put their right index finger on a touch-sensitive force-sensor (Piezosensor, PCB model 208C 02). The force sensor was positioned at the subject's elbow level (Figure 4.2).

The subjects keep their elbows at 90 degrees. In the LT condition, an oscilloscope

Figure 4.2: The experimental setup. The participant placed the right index finger inside of the aluminum frame and was able to push down. The frame was attached to the force sensor.

monitored the applied forces and emitted an auditory alarm when the threshold force of 1N was exceeded and then the trial was discarded and repeated. Only two cases were discarded; one each from an injured subject and an able-bodied control in light touch condition. In the HT condition, there was no limitation in the amount of finger force that could be applied. Each unique condition was tested two times and each trial lasted 30 s.

To examine the interaction between surface and touching, participants were tested under two different sitting surface conditions (i.e. stable condition (without flexible disk) and unstable condition (with flexible disk)). A 33 cm inflatable rubber disk (Fitness Gear, Inc, Coraopolis, PA) was established at unstable condition (Reeves et al., 2006).

Participants stared at a point at eye height on the wall. The wall was approximately 3m distance from the subject. Participants were instructed to sit still during each trial. Signals from the force plate were sampled at 100 Hz by a computer via an analog to digital interface

board.

4.2.3 Data analysis

Signals from the force plate were filtered with a 4th order low pass Butterworth filter with an adequate cut-off frequency. The adequate cut-off frequency of 5 Hz was measured with residual plot analysis (Winter, 2005). The center of pressure – a reflection of the system's neuromuscular response to the imbalances of the body's center of gravity – was separately calculated along with AP and ML axis by using following equations:

$$COP_{AP} = (-h \times F_x - M_y)/F_z$$
$$COP_{ML} = (-h \times F_y + M_x)/F_z$$

where h is the offset between the force plate sensors and the surface ($h = 58.7\,\text{mm}$) The same measures were used for calculating complexity as in Chapter 3.

Several traditional measures (mean sway velocity, sway ranges, and root mean square (RMS)) along the AP and ML direction were calculated (Prieto et al., 1996).

Mean velocity (mm/s) was calculated by the sum of the displacements of the COP at the given direction divided by total number of data point (n).

$$\text{Mean velocity} = \frac{1}{n-1}\sum_{i=1}^{n-1} \dot{\mathbf{r}}_{COP}(i)$$

Sway range (mm) was defined as the absolute value of the difference between the maximal and minimal values for the given direction.

$$\text{Sway range} = |COP_{\max} - COP_{\min}|$$

RMS (mm) from the mean COP is equivalent to the SD of the zero-mean adjusted COP

time series for the given direction.

$$\text{RMS=SD} = \sqrt{\frac{1}{n}\sum_{i=1}^{n-1} |\mathbf{r}_{COP}(i)|^2}$$

4.2.4 Statistical analysis

Each dependent variable was placed into a 3-way ($2 \times 3 \times 2$) mixed model ANOVA where group was a between factor, and surface (stable vs. unstable) and touch condition (no touch/ light touch/ heavy touch) were the within-subject factors. Separate ANOVAs with repeated measures and paired t-tests were followed to test significant interactions. While the α-level for significance was ≤ 0.05, a Holm-modified Bonferroni correction was applied to control for type-I error caused by multiple comparisons. Significance was set at $p < 0.05$.

4.3 Results

The mean finger tip force in each group are reported in Figure 4.3A. There were significant main effect of touch ($F(1, 14) = 158.33$, $p < 0.01$; $\eta_p^2 = 0.91$) and surface ($F(1, 14) = 7.53$, $p = 0.01$; $\eta_p^2 = 0.35$), but no main effect of group ($F(1, 14) = 0.73$, $p = 0.40$; $\eta_p^2 = 0.05$). Also there was interaction between touch and surface($F(1, 14) = 8.76$, $p = 0.01$; $\eta_p^2 = 0.38$). Post hoc analysis revealed that the heavy touch condition had larger force than the light touch and the no touch condition. The surface effect was found to result from there being greater in the unstable surface than stable surface condition.

4.3.1 Variability

The summaries of the 3 way mixed model ANOVA are shown in Table 4.1 and Table 4.2. For all variability measures, except for mean velocity along the AP axis, there were significant main effects of group, touch and surface. There was no main effect of surface on mean velocity

along the AP axis. Interactions between group and surface were significant for mean velocity along the AP and ML axes, and sway range along the AP axis. Also, interactions between group and touch for sway range along the ML axis, and interactions between surface and touch for mean velocity and RMS along the ML axis were found (Table 4.1).

The effects of group, surface and touch on postural variability are depicted in Figure 4.3. To test the interactions among group and surface, paired t-tests were conducted for mean velocity along the AP and ML axes, and sway range along the AP axis. It was found that the control group decreased mean velocity and sway range along the AP axis in the unstable condition compared to the stable condition while SCI group increased mean velocity and sway range in the unstable condition compared to the stable condition. In contrast to the results along the AP axis, the SCI group had faster velocity along the ML axis (3.95 ± 0.16 vs. 6.60 ± 0.64 mm/s, $t(1, 23) = -4.11$, $p < 0.01$) in the unstable condition compared to the stable condition, while there was no effect in the control group (3.54 ± 0.27 vs. 3.52 ± 0.19 mm/s, $t(1, 23) = 0.06$, $p = 0.95$) (Figure 4.3).

To test the interactions between groups and touch condition, a separate ANOVA was conducted for sway range along the ML axis. For the sway range along the ML axis, SCI group showed that in both light touch and heavy touch, sway range along the ML axis were reduced compared to no touch (no touch: 13.37 ± 3.30 mm; light touch: 9.34 ± 2.20 mm; heavy touch: 7.93 ± 1.29 mm; light touch vs. no touch ($p = 0.01$); heavy touch vs. no touch ($p = 0.03$)). However, controls had smaller sway range only in light touch than no touch condition (2.77 ± 0.307 vs. 3.80 ± 0.48 mm, $p < 0.01$) (Figure 4.3).

To test the interactions between surface and touch for sway range and RMS along the ML axis, a separate ANOVA was conducted. Light touch had a significant smaller sway range and RMS along the ML axis than heavy touch in the stable condition ($p = 0.02$ and $p = 0.01$ respectively). Light touch and heavy touch made significant smaller sway range along the ML axis and RMS along the ML axis than no touch in the unstable condition (Range ML: $p = 0.01$ and $p = 0.03$; RMS ML, $p = 0.01$ and $p = 0.03$ respectively) (Figure 4.3).

4.3.2 Complexity

The results of the complexity analysis are shown in Table 4.2 and depicted in Figure 4.4. There was a significant main effect of surface on SampEn along the AP axis. Interactions between group and surface were significant on SampEn along both AP and ML axis.

To test the interactions between group and surface, paired t-tests were conducted for SampEn of the COP trajectory along the AP and ML axes. Examination of SampEn along the AP axis revealed that there was no effect of surface condition in the SCI group ($t(1, 23) = -0.24$, $p = 0.80$), whereas controls had more complexity in stable condition than unstable condition (0.42 ± 0.02 vs. 0.25 ± 0.02, $t(1, 23) = 7.50$, $p < 0.01$). For SampEn along the ML axis, subjects with SCI had more complexity in the unstable condition than stable condition (0.25 ± 0.01 vs. 0.19 ± 0.01, $t(1, 23) = 0.01$) whereas controls had more complexity in the stable condition than unstable condition (0.30 ± 0.02 vs. 0.23 ± 0.01, $t(1, 23) = 3.19$, $p < 0.01$) (Figure 4.4).

Table 4.1: Summary of analysis of variance for six variability measures.

Independent variable	MV_{AP}			MV_{ML}			$Range_{AP}$			$Range_{ML}$			RMS_{AP}			RMS_{ML}		
Main effect	F-ratio	p-value	η_p^2	F-ratio	p-value	η_p^2	F-ratio	p-value	η_p^2	F-ratio	p-value	η_p^2	F-ratio	p-value	η_p^2	F-ratio	p-value	η_p^2
Group	7.15	**0.01**	**0.33**	9.64	$<$ **0.01**	**0.40**	14.55	$<$ **0.01**	**0.51**	14.58	$<$ **0.01**	**0.51**	13.43	$<$ **0.01**	**0.49**	14.15	$<$ **0.01**	**0.29**
Surface	2.27	0.154	0.14	6.98	**0.01**	**0.33**	10.71	$<$ **0.01**	**0.43**	5.75	**0.03**	**0.29**	9.70	$<$ **0.01**	**0.40**	18.88	$<$ **0.01**	**0.35**
Touch	4.02	**0.02**	**0.22**	6.45	$<$ **0.01**	**0.31**	5.48	**0.01**	**0.28**	7.43	$<$ **0.01**	**0.34**	5.49	**0.01**	**0.28**	10.62	$<$ **0.01**	0.23
Interaction																		
Group × surface	10.14	$<$ 0.01	**0.42**	7.12	0.01	**0.33**	10.71	$<$ 0.01	**0.15**	2.14	0.16	0.13	1.53	0.23	0.09	1.58	0.21	0.04
Group × touch	0.87	0.43	0.05	0.68	0.51	0.04	2.13	0.13	0.13	3.48	0.04	**0.19**	2.40	0.10	0.14	0.66	0.51	0.01
Surface × touch	3.13	0.05	0.18	4.10	0.02	**0.22**	2.89	0.07	0.17	3.33	0.05	**0.19**	2.53	0.09	0.15	5.36	0.00	**0.13**
Group × surface × touch	3.08	0.06	0.18	2.97	0.06	0.17	0.95	0.39	0.06	1.20	0.31	0.07	0.91	0.41	0.06	1.94	0.15	0.05

Table 4.2: Summary of analysis of variance for two complexity measures.

Independent variable	$SampEn_{AP}$			$SampEn_{ML}$		
Main effect	F-ratio	p-value	η_p^2	F-ratio	p-value	η_p^2
Group	2.35	0.14	0.14	2.14	0.16	0.01
Surface	15.98	$<$ **0.01**	**0.53**	0.02	0.88	0.03
Touch	0.12	0.88	0.01	0.33	0.72	0.04
Interaction						
Group × surface	19.27	$<$ **0.01**	**0.57**	15.33	$<$ **0.01**	**0.25**
Group × touch	2.12	0.13	0.13	1.42	0.25	0.07
Surface × touch	0.98	0.38	0.06	0.51	0.60	0.17
Group × surface × touch	1.46	0.24	0.09	1.10	0.34	0.02

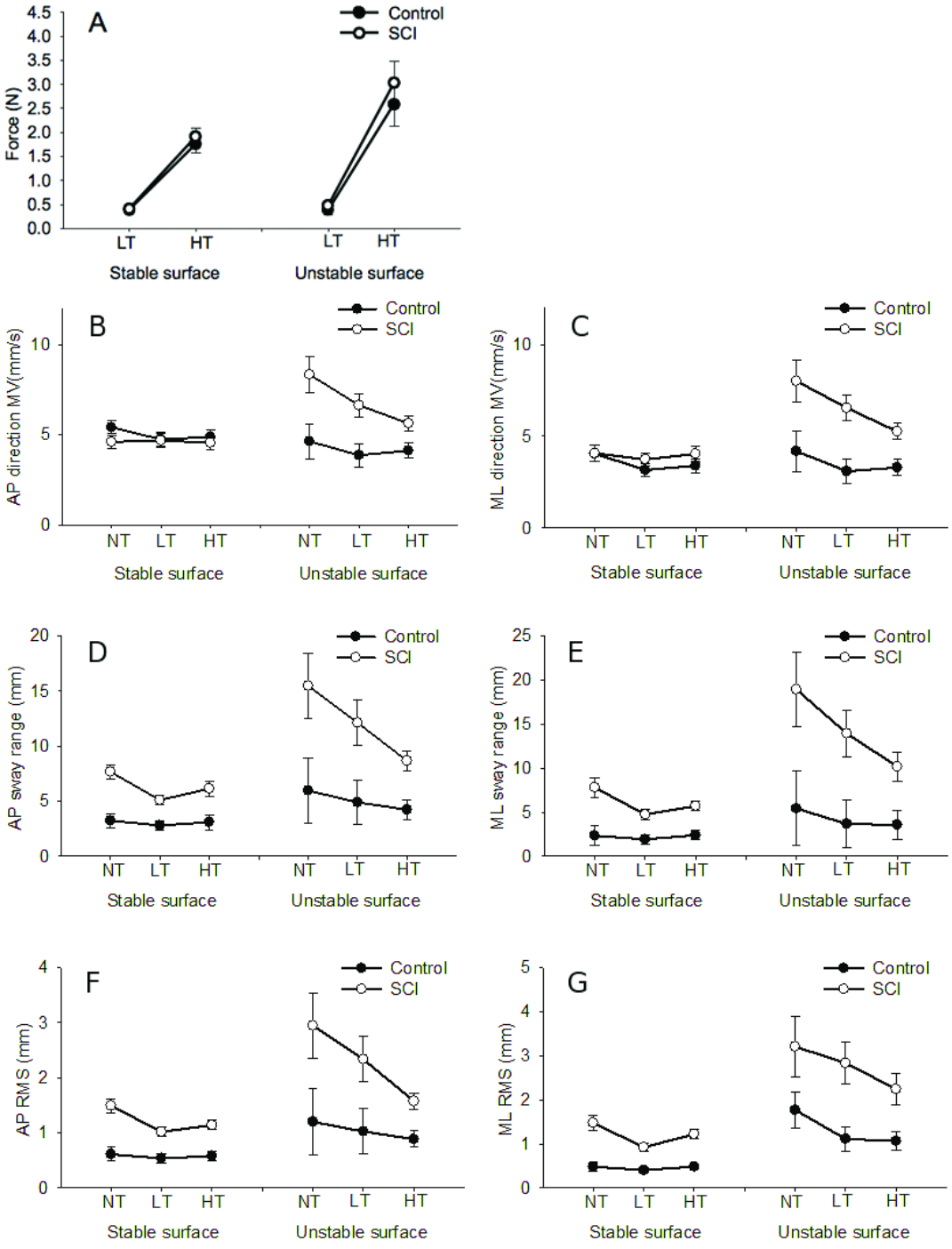

Figure 4.3: Plots showing mean and SE for traditional variability measures in both stable condition and unstable condition: A) mean cutaneous force B) AP direction mean velocity (mm/s) C) ML direction mean velocity (mm/s), D) AP direction sway range, E) ML direction sway range, F) AP direction RMS, and G) ML direction RMS.

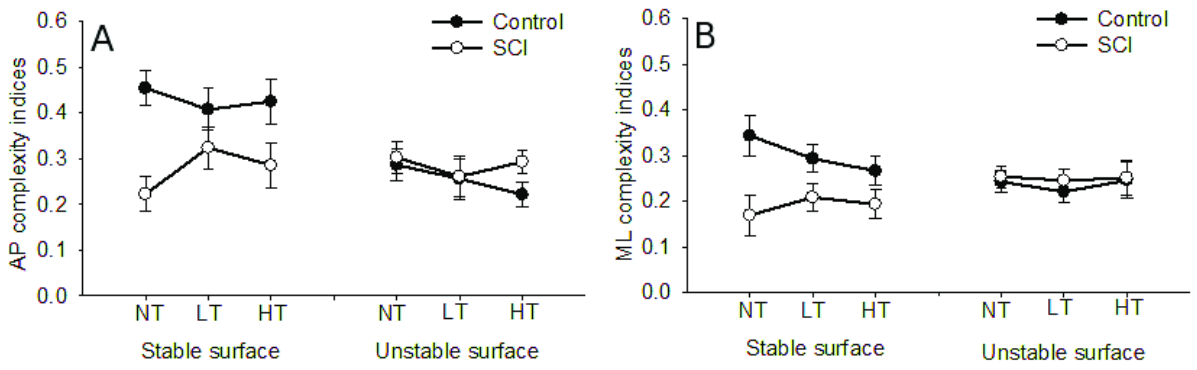

Figure 4.4: Plot showing mean and SE for SampEn in both stable and unstable conditions: A) SampEn in AP direction, B) SampEn in ML direction.

4.4 Discussion

This investigation examined the effect of additional cutaneous information on the variability and complexity of sitting postural control of persons with SCI and controls. Congruent with our hypothesis, there was a decrease in variability of COP trajectory with the additional cutaneous information in the SCI group, while there was a minimal effect in the control group. However, counter to our hypothesis, complexity of seated postural control did not change with additional sensory information.

Additive Sensory Information and Postural Variability

This is the first investigation examining the effect of additional sensory information on seated postural control in persons with SCI. Overall; it was found that the effect of touch was greater in persons with SCI than healthy controls. Indeed, there was no effect of additional sensory information on the control's seated postural output. This is congruent with previous research that has documented that additional sensory information is most effective in groups with reduced postural control such as older adults (Baccini et al., 2007), individuals with anterior cruciate ligament injury (Bonfim et al., 2008), and persons with diabetic neuropathy (Dickstein, Peterka, & Horak, 2003). It was theorized that available sensory information

is automatically utilized to adapt to changing environment condition (Mahboobin, Loughlin, Atkeson, & Redfern, 2009). The primary source for information about body orientation is proprioception in non-injured persons. However, persons with SCI have reduced proprioception. Therefore, deficits in proprioception experienced by persons with SCI could be compensated by other available sources such as touch (Peterka, 2002). This dynamic regulation of sensory sources, which is so called sensory reweighting, is more helpful to increase postural stability in challenging situations, e.g. unstable condition (Creath et al., 2002).

Our observations confirm that persons with SCI can utilize sensory information concerning their postural control. This raises the possibility that biofeedback could be used to augment seated postural control in persons with SCI (Huang, Wolf, & He, 2006) and potentially reduce falls. This potential benefit is key give that falls from sitting postures are common and often injurious in persons with SCI (Berg, Hines, & Allen, 2002; Boswell-Ruys, Harvey, Delbaere, & Lord, 2010). Moreover, there is limited research on techniques to minimize falls in persons with SCI.

4.4.1 Additive Sensory Information and Postural Complexity

Incongruent with our hypothesis, touch had no effect on the dynamic structure (i.e. complexity) of postural control. This differential change in variability and complexity metrics is further support that these metrics are indicative of distinct properties of the observed motor behavior (Sosnoff & Newell, 2006). A potential reason for the lack of change in postural complexity with additive sensory input is that persons with SCI could not make appropriate motor responses with the augmented sensory input.

Interestingly, in the surface condition, complexity of sitting posture of persons with SCI is the opposite trend compared to persons without SCI. For example, when we compared the two groups, in the stable condition, COP movement of persons with SCI is less complex as than controls. However, in the unstable condition, COP movement of persons with SCI is more complex than the controls. The effect of surface on the complexity of seated postural

control is congruent with the loss of adaptability hypothesis (Vaillancourt & Newell, 2002; Vaillancourt et al., 2004) as discussed in Chapter 3. Especially along the ML axis, there were inverse relationships between surface and group. One interpretation of this similar complexity level in unstable condition between persons with SCI and without SCI is that it stems from different postural strategies. It is possible that persons with SCI utilized an active strategy characterized by more movement adjustments to compensate for limited postural muscle activity in the unstable condition. However, persons without SCI used a passive strategy characterized by postural muscle co-contraction to maintain their posture with intentional restricted movement.

4.5 Limitations of study

This investigation had several limitations. First, minimal practice with the touch condition was provided to participants. Additionally, the order of experimental conditions was not randomized. The no touch condition was performed first, followed by the light touch then the heavy touch condition. Although unlikely, it is possible that group and condition differences stem from differential learning. Lastly, in the unstable condition, the height of rubber disk which needs to be taken into account when determining COP, was affected by the weight of subjects (i.e. no uniform across participants nor trials). We used an average value of the disk height (i.e. about 3.81 cm) in calculating COP. The use of an average value could potentially influence the measures of COP amplitude and sway. However, it would not have any effect on structural measures.

4.6 Conclusion

Additional sensory information from cutaneous sense reduces the variability, but not the complexity of the sitting balance in persons with SCI. The beneficial effect of additive sensory information was greatest for persons with SCI than persons without SCI. The findings raise the idea that biofeedback could be a useful technique to minimize fall risk in persons with SCI.

Chapter 5

Spinal cord injury and time to instability in seated posture

Abstract

Seated postural control is an essential motor task for individuals with SCI. Although individuals with SCI maintain a seated posture for an extend time during daily physical activities, they have decreased seated postural control compared to able-bodied counterparts. There are numerous methods to examine postural stability. One method to determine postural stability is by calculating the virtual time to contact (VTC) to the stability boundary (Slobounov et al., 1997). The virtual time provides an estimate of the time required for the center of pressure (COP) to reach the functional stability boundary. It is calculated at every point in time series of the COP.

Importantly, this method does not measure the relative position of the COP to stability boundary, but rather estimates the time needed to reach to the boundary (Slobounov, Haibach, & Newell, 2006). As such it does not require losses of stability, making it ideal for clinical populations. In the current investigation, we examine VTC in SCI and controls in order to better understand seated postural control of SCI individuals. We expect SCI individuals to have smaller VTC compared to individuals without SCI. Additionally we predict that individuals with higher injury level (i.e. lower function) would have smaller VTC compared to those with lower injury levels. It is also predicted that biomechanical constraints (i.e. foot support) and sensory information (i.e. vision) will influence VTC.

Methods

A total of 36 young adults (7 high spinal cord injured (HI), 11 low spinal cord injured (LI), and 18 non-spinal cord injured persons who are age, gender, and sitting matched) had their seated postural control recorded. The experimental setup is the same as described in Chapter 3. In order to calculate VTC, COP data from the quiet sitting condition of Chapter 3 were utilized. Additionally, the functional stability boundary of each subject was determined by having them lean forward, backward, laterally and diagonally pivoting at the hip joint, in the circular direction leaning as far as possible without losing balance for one minute. To examine the effects of biomechanical constraints (i.e. foot support) and vision on functional boundary and VTC, subjects conducted four static postural tasks: sitting still with eyes open (EO) and eyes closed (EC) with and without foot support.

Results

Analysis of VTC revealed that HI and LI group had smaller functional boundary than the control group. The HI group had shorter mean VTC than the control group. The HI had larger complexity index of VTC than the control group. Also, there was an interaction between group, vision and support condition on complexity of VTC. In supported eyes closed condition, the HI group had more complex VTC than LI group and the control group, However, in unsupported eyes closed condition, there was no difference among groups.

Conclusion

The differential effect of SCI, vision and biomechanical constraints on VTC highlights that this approach is worthwhile to quantify seated postural control in different populations. The HI group had less of a temporal safety margin as indexed by a shorter VTC. This observation suggests that individuals with higher levels of injury are more prone to falls. Nevertheless, HI group use more active strategy to prevent fall than controls and LI group. Support

and vision change the dynamic structure of VTC especially in persons with higher injury. Further work examining ways to minimize instability in seated postural control in persons with SCI is warranted.

5.1 Introduction

Postural control is a complex output of the neuromuscular system resulting from the interaction of multiple control processes using multi-sensory information from vision, vestibular, proprioception systems (Gagnon et al., 2005). Degeneration or loss of function in any of these subsystems can lead to deficits of postural control.

For individuals with spinal cord injury (SCI), seated postural control is one of the most fundamental daily activities because they usually maintain a sitting posture for an extended time. Persons with SCI control seated posture differently from that of non-injured persons (Seelen et al., 1997). The different seated postural control is due to multiple factors such as modified processing of sensory information in the CNS; atrophy and scoliosis of the erector spinae, (a bundle of postural muscles and tendons of the back) and reduced proprioception. It is often assumed based on sway metrics that persons with SCI have a less stable seated posture than able bodied persons (Seelen et al., 1997, 2001). However, the evidence to support this notion is ambiguous.

Concerns of sitting instability are warranted, given that falls from sitting postures are very common in people with SCI. 40% of SCI individuals in the U.S. experienced falls, and 47% of them suffered a fall-related injury (Berg et al., 2002; Boswell-Ruys et al., 2010). Falling is an example of loss of stability. Postural stability normally has been evaluated with variability of center of pressure (COP) – a reflection of the system's neuromuscular response to the imbalances of the body's center of gravity (Winter, 2005; Duarte & Freitas, 2010). The majority of research related to the sitting posture in persons with SCI has examined standard statistics of COP motion (e.g. mean, standard deviation (SD), coefficient of variance (CV),

etc) (Cavanaugh, Guskiewicz, & Stergiou, 2005). However, standard statistics are not able to access how well someone can maintain their posture within their stability boundary. That is, standard COP analyses do not provide a direct index of postural stability (Slobounov et al., 2006).

Determination of the virtual time to contact to the stability boundary has been proposed as a direct measure of postural instability (Slobounov et al., 1997). Postural instability occurs when the center of pressure moves outside of a stability boundary (Slobounov et al., 2006). The term "virtual" means that the individuals control their posture not to actually contact with the stability boundary. As such it does not require losses of stability making it ideal for clinical populations, such as persons with SCI.

Another advantage of VTC approach is that it incorporates the spatial and temporal features of postural sway which provide a more topological quantification of an individual's postural strategy. The VTC approach takes into account acceleration, velocity, and position of the COP trajectory to estimate the temporal margin to the stability boundary (Slobounov et al., 1998). Examinations of VTC have revealed that older adults (Slobounov, Moss, Slobounova, & Newell, 1998) and persons with Parkinson's disease (Wegen, Emmerik, Wagenaar, & Ellis, 2001), and persons with multiple sclerosis (Gruber et al., 2011; Cattaneo, Ferrarin, Jonsdottir, Montesano, & Bove, 2012) have a smaller VTC (e.g. time to losing stability) while maintaining an upright stance than control participants. It is maintained that the reduced VTC in these clinical populations is functionally relevant because it suggests that they have less time to recover from a postural perturbation and consequently at a greater chance of falling. However, there has been no examination of seated postural control utilizing this functionally relevant metric.

Stability of posture is well known to be influenced by the availability of sensory information. For instance, VTC is decreased in various populations when individuals maintain a posture with their eyes closed compared to eyes open (Slobounov et al., 1998). The availability of foot support has been suggested to increase stability in seated postural control (Hong

et al., 2008; Janssen-Potten, Seelen, Drukker, Spaans, & Drost, 2002). Despite the logic in these reports, there is minimal evidence that visual information and/or foot support actually increases seated postural stability in persons with SCI. Consequently, it was of interest to determine how visual information and biomechanical support influenced the stability as indexed by VTC of postural control in persons with SCI compared to controls.

Despite the advantages of the VTC to traditional approaches, this statistic does not quantify the dynamic properties (e.g. complexity) of postural control. An approach to quantify the dynamic properties of postural control is to determine the COP complexity. Various metrics such as multiple scale entropy (MSE), detrended fluctuation analysis (DFA) and fractal dimension have been shown to be sensitive to changes in dynamic properties of postural control due to aging, disease or injury (Stergiou, 2004). It is possible that the dynamical properties of the VTC time series will provide unique information concerning postural control in persons with SCI.

It was expected that through the enhanced sensitivity of the VTC measures, different strategies of postural control between non-SCI and SCI subjects, which are not detectable using conventional COP-based measures, should become clear. The time series of VTC would be a better way to examine different characteristics of postural complexity in that it can explain the strategy of postural control. This approach could more directly explain how persons actually control their body movement to maintain their body within the stability boundary to avoid the fall.

Finally, the purpose of this investigation is to determine whether the VTC during sitting is capable of quantifying postural instability in persons with SCI. It was predicted that individuals with SCI would have smaller VTC compared to healthy controls; VTC would scale with injury level in persons with SCI.

5.2 Methods

5.2.1 Participants

36 persons (18 persons with SCI and 18 non-spinal cord injured persons who are age, gender, and sitting matched individuals (control group)) participated in this study. The SCI group was divided into a high injury (HI) group ($n = 7$; SCI above T10) and a low injury group ($n = 11$; SCI between T11 to L4). Experimental procedures were approved by the local institutional review committee and all participants provided written informed consent at the beginning of the experiment.

5.2.2 Procedures

To quantify seated postural control, participants sat on a force platform (AMTI, Inc.) on a custom wooden box (1.5 m $\times$ 0.75 m $\times$ 0.75 m) with their arms by their side for 30s. In order to calculate VTC, the functional stability boundary was determined by the procedures of Slobounov et al. (1997). The functional stability boundary of each subject was determined by having them lean forward, backward, laterally and diagonally pivoting at the hip joints to trace a circle while leaning as far as possible without losing balance for one minute (Slobounov et al., 1997). Two vision conditions (eyes open and eyes closed) and two support conditions (supported and unsupported) were combined into four separate conditions.

5.2.3 Data Analysis

The center of pressure – a reflection of the neuromuscular response to the imbalances of the body's center of gravity (Winter, 2005) was calculated as:

$$COP_{AP} = (-h \times F_x - M_y)/F_z$$

$$COP_{ML} = (-h \times F_y + M_x)/F_z$$

where h is the offset between the force plate sensors and the surface ($h = 20.6\,\text{mm}$).

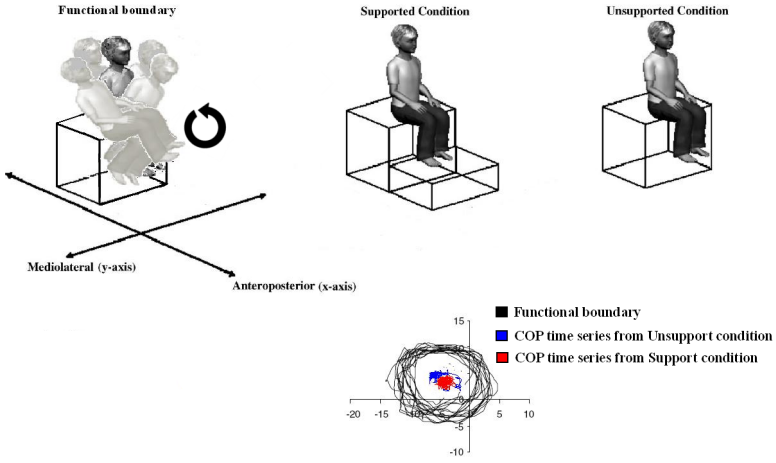

Figure 5.1: Illustration of the experimental setup, and example of functional boundary and COP time series; modified from Hong et al. (2008).

The functional boundary was calculated by a direct least square fitting method (Fitzgibbon, Pilu, & Fisher, 1999). In Figure 5.3, the red line illustrates the actual ellipse fitted to the individuals' functional boundary while they pivoted. The method led to calculate the functional area with high robustness to noise.

For the VTC calculation, a position vector of the COP on a virtual trajectory $\tau_i(t)$ was determined for each moment in time t_i based on the instantaneous COP velocity and acceleration:

$$x, y(\tau) = r_{x_i,y_i}(t_i) + v_{x_i,y_i}(t_i)\tau + a_{x_i,y_i}(t_i)\tau^2/2$$

where $r_{x_i,y_i}(t_i)$ is the instantaneous position vector, $v_{x_i,y_i}(t_i)$ is the instantaneous velocity vector and $a_{x_i,y_i}(t_i)$ is the instantaneous acceleration vector in the x and y directions.

With the current virtual trajectory, the position vector for crossing point (x_c, y_c) were determined by

$$\left(\frac{x_c}{R_x}\right)^2 + \left(\frac{y_c}{R_y}\right)^2 = 1$$

where, as in Chapter 2, R_x and R_y are the semiaxes of the ellipse traced out by the initial pivoting.

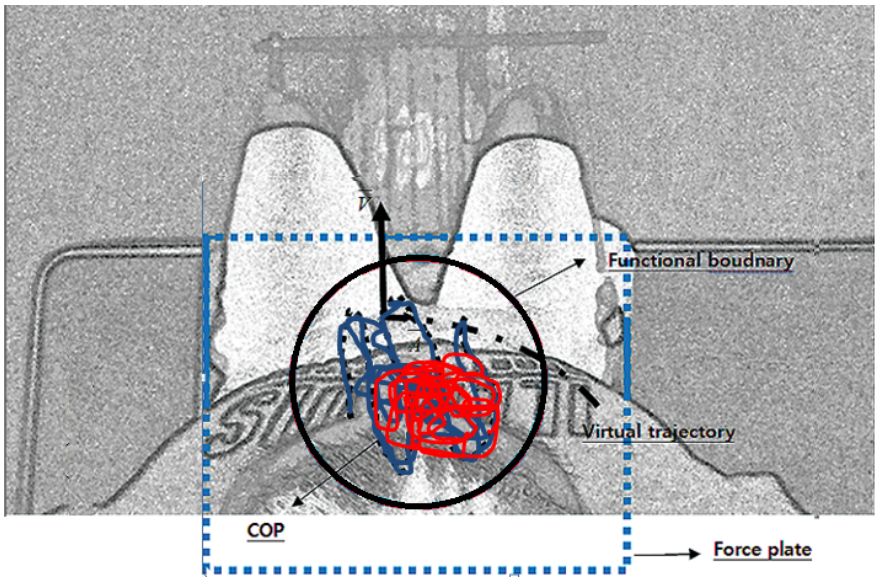

Figure 5.2: Schematic of VTC computation with respect to two-dimensional functional stability boundary during sitting still.

The same measures were used for calculating complexity of VTC as in Chapter 3. Adaptability index was defined as difference between SampEn COP in sitting still and SampEn COP during voluntary rocking which were calculated in Chapter 3.

5.2.4 Statistical analysis

Each dependent variable was placed into a 3-way ($3 \times 2 \times 2$) mixed model ANOVA where group was a between factor, and support (supported vs. unsupported) and vision condition (eyes open vs. eyes closed) were within subject factors. Separate ANOVAs with repeated measures 2×3 (group and support) for each vision condition was followed to test interactions. Dependent t-test was followed to test interaction between group and support for each group. To determine the association between VTC and adaptability, Spearman rho correlation was conducted in a subset of the data. All data were analyzed using SPSS version 18.0 (SPSS Inc., Chicago, IL., USA). Significance was set at $p < 0.05$.

5.3 Results

Figure 5.3 showed representative trials of the COP for the boundary trials and EO sitting trials from a subject of each group. Overall, HI group showed smaller functional boundary but larger COP sway area than LI group and control group.

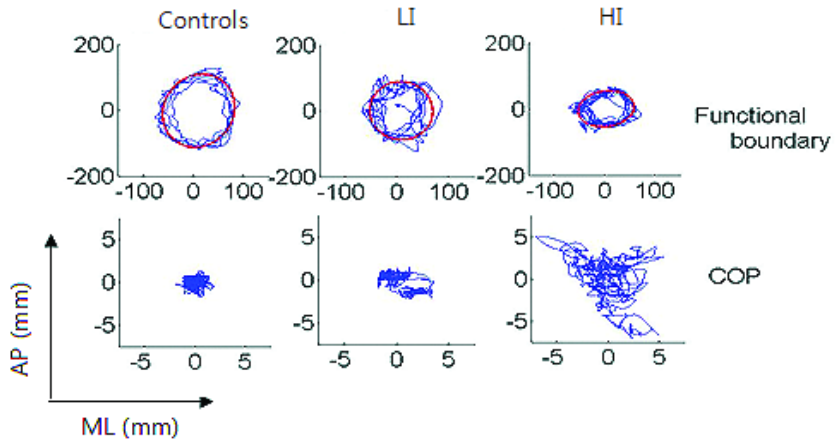

Figure 5.3: Representative trials of the COP for the boundary trials and EO sitting trials from a subject of each group. The red line depicts the functional boundary (top) and the blue trace represents the COP trajectory (bottom).

Examination of functional boundary revealed that the HI group ($p < 0.01$) and LI group ($p = 0.04$) had smaller functional boundary than the control group (174.87 vs. 118.21 vs. 73.3 mm^2). However, there were no difference between LI and HI groups ($p = 0.24$) (Figure 5.4).

The summary of the 3-way mixed model ANOVA is shown in Table 5.1. For the mean VTC, there was a significant main effect only for group. Overall, HI group shorter VTC than controls (0.23 vs. 0.28, $p = 0.01$). However there were no difference between LI group and controls ($p = 0.23$) and HI group and LI group ($p = 0.10$). Also there is an interaction between group and support. To examine the interaction between group and support in mean VTC, dependent t-tests were conducted separately for each group. Overall, it was found that only the control group's mean VTC was influenced by support condition (Unsupport (0.28)

vs. support (0.29); $t(35) = 2.98$, $p < 0.01$) (Figure 5.5).

For the SampEn of VTC, there were no significant main effects of group, or vision support. However, there were significant interactions between group and support, between support and vision, and among group, support, and vision. However, for the SampEn of COP, there were no main effects or interactions (See Table 5.1 and Figure 5.6).

Table 5.1: Summary of analysis of variance for each independent variable.

Independent variable	Mean VTC			SampEn of VTC			SampEn of COP		
Main effect	F-ratio	p-value	η_p^2	F-ratio	p-value	η_p^2	F-ratio	p-value	η_p^2
Group	4.16	**0.02**	0.20	2.75	0.07	0.14	1.41	0.25	0.06
Surface	0.02	0.89	< 0.01	3.18	0.08	0.08	0.12	0.72	< 0.01
Touch	1.99	0.16	0.05	0.06	0.79	< 0.01	1.15	0.29	0.01
Interaction									
Group × surface	3.91	**0.03**	0.19	7.68	$<$ **0.01**	0.32	0.73	0.48	0.05
Group × touch	2.08	0.14	0.11	1.71	0.19	0.09	2.87	0.07	0.13
Surface × touch	0.07	0.78	< 0.01	5.33	**0.04**	0.11	0.35	0.55	< 0.01
Group × surface × touch	0.21	0.81	0.01	3.78	**0.03**	0.18	2.84	0.10	0.14

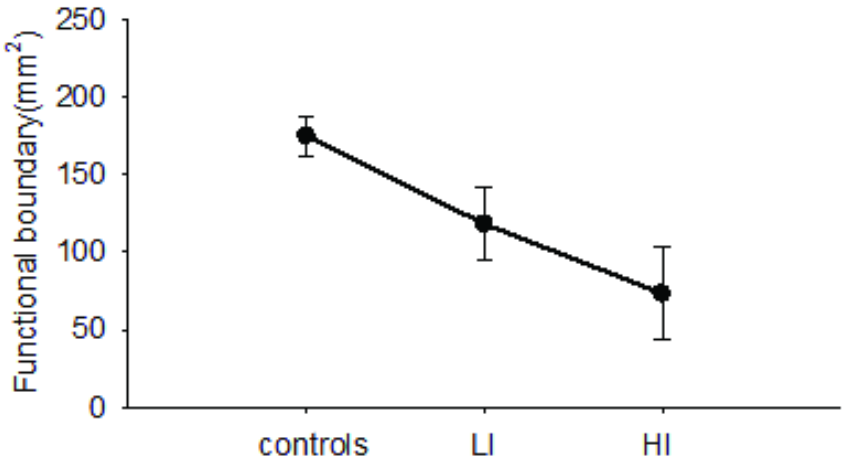

Figure 5.4: Comparisons of mean functional boundaries.

Overall, HI group Overall, HI group had larger SampEn of VTC than controls (1.86 ± 0.05 vs. 1.71 ± 0.03, $p = 0.02$). To examine the interaction among group, vision and support in SampEn of VTC, a separate ANOVA was conducted for each vision conditions. At the eyes open condition, there was no main effect for group ($F(1, 33) = 1.39$, $p = 0.26$) support ($F(1, 33) = 0.00$, $p = 0.98$) and interaction between group and support ($F(2, 34) = 1.70$,

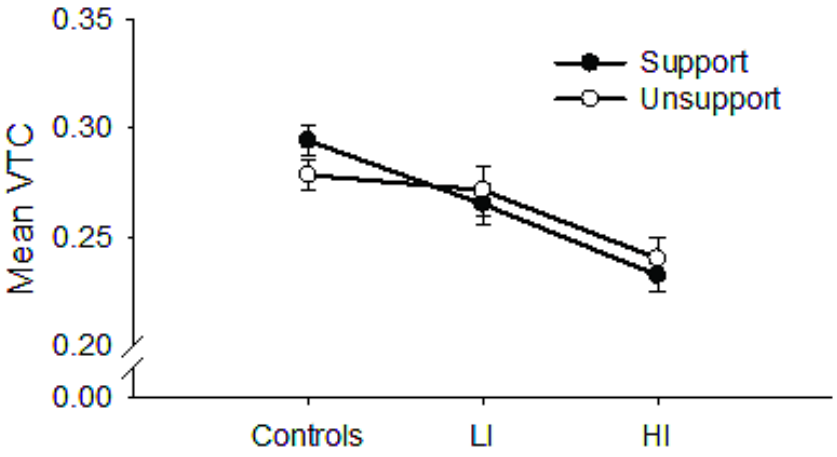

Figure 5.5: Comparisons of mean VTCs to the stability boundary.

$p = 0.19$). However, in the eyes closed condition, there were main effect for group ($F(2.33) = 4.89$, $p = 0.01$) and support ($F(1, 33) = 5.53$, $p = 0.02$), and an interaction between group and support ($F(2, 33) = 8.25$, $p < 0.01$). Separate ANOVA revealed that in supported eyes closed condition, the HI group had larger SampEn of VTC than LI group and the control group (2.11 (HI) vs. 1.77 (LI) vs. 1.67 (controls), $p = 0.01$ and $p < 0.01$ respectively.)

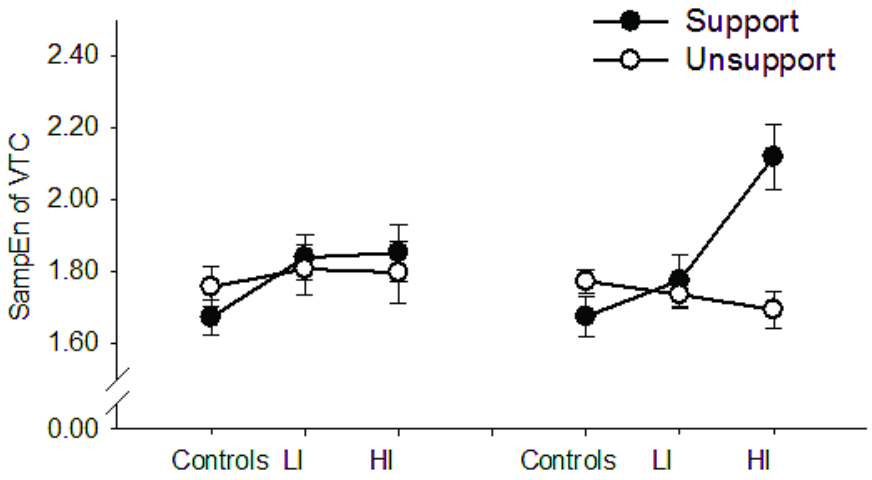

Figure 5.6: Comparisons of SampEn of VTC as a function of support, vision (eyes open/closed) and group.

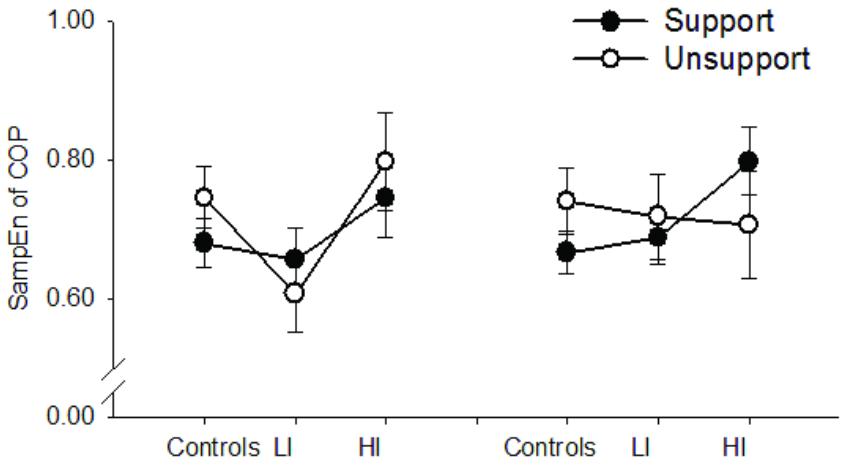

Figure 5.7: Comparisons of SampEn of COP as a function of support, vision (eyes open/closed) and group.

5.4 Discussion

The purpose of this investigation was to determine whether the VTC during sitting is capable of quantifying postural instability in persons with SCI. To achieve this purpose, we examined if the individuals with SCI have different virtual time to contact (VTC) to the functional stability boundary and if there are different dynamic properties (i.e. complexity) in VTC of persons with SCI compared to persons without SCI. The novel findings of this investigation were that persons with SCI had a smaller stability boundary than persons without SCI and that there was lower mean VTC in the SCI groups. However, complexity of VTC was larger in persons with SCI.

Individuals with spinal cord injury at T10 and above lack control of trunk musculature (Seelen et al., 1997). It is most likely that the lack of control of the trunk musculature plays a dominant role in the HI group's reduced functional boundary. The LI group which has trunk musculature control also demonstrated a smaller functional stability boundary than controls. This observation indicates that other factors, in addition to trunk musculature control contribute to the functional stability boundary. For instance, it is possible that reduced sensory inputs cause individuals to be more cautious in their estimation of their

stability boundary. Future work could determine the association to individual's perceived functional boundary and the actual functional boundary.

There were no significant differences in functional boundary between HI and LI. The lack of difference in functional stability boundary between HI and LI should not be interpreted as evidence that they have the same postural control stability. In fact, the examination of virtual time to contact revealed that the HI group had smaller virtual time to contact than the LI and control group. This observation supports that persons with higher level of injury are less stable and possibly at greater risk of falls. It also highlights the importance of viewing the functional boundary along with the VTC metric. In isolation the functional boundary can provide misleading and/or incorrect information.

There were significant differences in complexity in VTC as a function of group and condition. However, these group differences in complexity were in contrast to our hypothesis, which was that there would be a reduction in complexity of VTC in the SCI groups compared to the controls. The HI group had the greatest complexity of VTC time series. A greater amount of complexity is indicative of greater active control (Lipsitz & Goldberger, 1992). It is possible that individuals with higher injury level utilize more active modulations of their seated postural control to maintain stability. Support for this interpretation stems from the observation that the highest level of complexity is observed in the HI group in the eyes closed with foot support condition. An alternative interpretation of the data is that the greater complexity of HI group could be interpreted as a indication an indication of less control, such that the system is uncontrolled that the body sways everywhere. Further work is necessary to delineate between these contrasting interpretations.

It is in this condition that they would be able to have the greatest amount of active control. There are several ways to test this possibility. Future work could utilize muscular recordings (e.g. EMG) during seated postural control tasks and determine if a greater activation of musculature is necessary in persons with higher injury compared to other groups. Alternatively, control participants could be instructed and/or trained to maintain a

seated posture with non-postural musculature and a direct comparison of the complexity of the VTC could be conducted.

The higher amount of complexity in the LI group is also in contrast to previous literature examining postural control complexity among the population with disease or disorders (Bolbecker et al., 2011; Gruber et al., 2011). For example, Gruber et al. (2011) demonstrated that there is decreased complexity of mediolateral COP in adolescent idiopathic scoliosis patients compared to healthy persons. There are several reasons for this discrepancy. First, complexity of VTC is independent of complexity of COP with each of the time series yielding different information. The majority of the studies examining COP complexity focus on movement along AP and ML axes separately. However, the VTC examines postural control in 2-dimensional space. Additionally, the VTC time series has functional relevance, while COP in of itself has less functional relevance.

5.5 Limitation of study

Despite the novel observations of this investigations there were several limitations. First, the majority of participants with SCI were wheelchair athletes. Consequently, generalization to the general SCI population is suspect. It is important to note that the VTC was still reduced in these athletes and it is logical to speculate that group differences would be greater if non-wheelchair athletes were tested. Another limitation is the calculation of the VTC is based on self-perceived limits of stability. It is possible that participants under estimate their functional stability boundary, which results in a reduced VTC. This is especially true in participants with SCI given the reduced proprioceptive function and potentially more severe consequences of losing stability. Further work examining the association between perceived limits of stability and actual limits of stability in persons with SCI is warranted.

5.6 Conclusion

Finally, VTC during sitting is capable of quantifying postural instability in persons with SCI. Specifically, differences in VTC as a function of SCI was found in both average and complexity measures of VTC. This study showed that VTC is applicable to seated postural control and provides a novel method to evaluate sitting balance. The stability of sitting of individuals with SCI is typically determined by measures of the amount of variability in the center of pressure or more particularly the position of the center of pressure relative to the stability boundary of sitting. Persons with SCI sitting stability was determined not only by movement pattern of the COP, but also by stability boundary which is determined by individual differences such as range of motion of upper body, flexibility, injury level, etc. Based on the functional relevance of VTC, it provides a novel metric to examine the effectiveness of various rehabilitation approaches and/or technologies aimed at improving seated postural control in persons with SCI. The ultimate test of the utility of VTC of seated postural control is to determine if VTC is truly related to falls in persons with SCI. This study has the potential to lead to a paradigm shift in research of seated postural control matching the functional relevance of clinical tests with the precision of force platform investigations.

Chapter 6

Conclusion

The current study was conducted to further understand seated postural control of individuals with SCI as a function of manipulations of sensory information (vision and proprioception), and task constraints (sitting still and body rocking). The overall purpose was to examine the predictions of the loss of complexity hypothesis and the loss of adaptability hypothesis on seated postural control of persons with and without SCI. It was generally hypothesized that seated postural control of individuals with SCI would be different by SCI level, sensory information (vision and proprioception) and task constraints compared to individuals without SCI.

A series of four experiments was conducted to test hypothesis, the conclusions were as follows: 1) In Chapter 3, the observation that the complexity of seated postural control in persons with SCI is dependent on the intrinsic dynamics of the task is congruent with the loss of adaptability hypothesis. The current observations highlight the limitations of the loss of complexity theory in neuromuscular output.

2) In Chapter 4, additional cutaneous sensory information reduces the variability, but not complexity of the sitting balance in persons with SCI. This observation highlights the difference in variability and complexity metrics. This finding indicates that additional cutaneous information is beneficial in persons not only in standing postural control but also in seated postural control. The ecological benefit of this observation remains to be determined.

3) In Chapter 5, the observation suggests that individuals with higher levels of injury are more prone to falls. Nevertheless, persons with SCI use more active strategy to prevent falls than low injured or non-injured persons. Support and vision change the dynamic structure

of VTC especially in persons with higher injury. Further work examining ways to minimize instability in seated postural control in persons with SCI is warranted.

Despite the novel observations, there were several limitations with this research project. First, each investigation was not necessarily distinct. Although the hypotheses addressed in each investigation were independent, data from various conditions (e.g. sitting still) was used across investigations. This potentially violates various statistical assumptions of independence and normality. No statistical approaches were utilized to account for these violations. The "sharing" of data was necessary due to the small number of participants with SCI available for testing. Secondly, as previously mentioned the majority of the participants with SCI were wheelchair athletes. Consequently, the generalization of the findings to the general SCI population is suspect. Third, in an attempt to reduce fatigue conditions were not randomized. This potentially resulted in a differential effect of fatigue across various experimental conditions.

Finally, the observations of the present research project leads to the general conclusion that even though persons with SCI had reduced postural control which were indexed by increased sway area, reduced dynamic structure, and shorter VTC, they actively use additional sensory input and develop different postural strategies to maintain stability. Further research examining the association between stability and falls in persons with SCI is warranted. The findings of this study suggests that in further examinations concerned with seated postural control in persons with SCI, new analytical techniques (i.e. VTC) and biofeedback approaches should be considered.

Appendix A

IRB consent forms

UNIVERSITY OF ILLINOIS
AT URBANA-CHAMPAIGN

Office of the Vice Chancellor for Research
Institutional Review Board
528 East Green Street
Suite 203
Champaign, IL 61820

September 1, 2010

Jacob Sosnoff
Kinesiology & Community Health
207 Freer Hall
M/C 052

RE: *Seated postural control in individuals with spinal cord injury*
IRB Protocol Number: 09361

Dear Jacob:

Thank you very much for forwarding the modifications to the University of Illinois at Urbana-Champaign Institutional Review Board (IRB) office for your project entitled *Seated postural control in individuals with spinal cord injury.* I will officially note for the record that these major modifications to the original project, as noted in your correspondence received May 27, 2010, adding a balance task and a wheelie task, and adding an additional investigator, have been approved. The expiration date for this IRB protocol, UIUC number 09361, is 02/04/2011. The risk designation applied to your project is *no more than minimal risk.*

As your modifications involved changes to consent forms, I am enclosing the revised forms with date-stamp approval. Please note that copies of date-stamped consent forms must be used in obtaining informed consent. If modification of the consent form is needed, please submit the revised consent form for IRB review and approval. Upon approval, a date-stamped copy will be returned to you for your use.

Please note that additional modifications to your project need to be submitted to the IRB for review and approval before the modifications are initiated. To submit modifications to your protocol, please complete the IRB Research Amendment Form (see http://irb.illinois.edu/?q=forms-and-instructions/research-amendments.html). Unless modifications are made to this project, no further submittals are required to the IRB.

We appreciate your conscientious adherence to the requirements of human subject research. If you have any questions about the IRB process, or if you need assistance at any time, please feel free to contact me or the IRB Office, or visit our Web site at http://www.irb.illinois.edu.

Sincerely,

Sue Keehn, Director, Institutional Review Board

Enclosure(s)

c: Sung-hoon Shin

telephone (217) 333-2670 • *fax* (217) 333-0405 • *email* IRB@illinois.edu

UNIVERSITY OF ILLINOIS
AT URBANA-CHAMPAIGN

Department of Kinesiology and Community Health

Louise Freer Hall
906 South Goodwin Avenue
Urbana, IL 61801-3895

Date _____

Adult Informed Consent Agreement

Project title: Seated postural control in individuals with spinal cord injury (SCI)

Purposes of the study:
The main purpose of the present study is to assess seated postural control in individuals with spinal cord injury. This project is conducted under the guidance of JJ Sosnoff in the Department of Kinesiology and Community Health at the University of Illinois at Urbana-Champaign. This research may enable a better understanding of postural control in individuals with SCI. It is anticipated that the data will be presented at scientific conferences and published in scientific journals.

What you will do:
During your visit you will be asked to complete a seated postural control task. You will be asked to sit on a custom wooden box placed on a force platform that measures your postural control. Some of the times that you are sitting on the platform you will be asked to balance on a Dyna Disk or perform a wheelie. Your wheelie task will be recorded by three digital cameras. Only your arms and trunk will be recorded.

Small reflective markers (5 mm diameter) will be placed on your arm and torso (Left shoulder, right shoulder, left anterior superior iliac spine, right anterior superior iliac spine, right lateral upperarm, right lateral elbow, right medial elbow, right lateral forearm, right lateral wrist, right medial wrist, right index finger, C7). Markers will be placed on clothing overlying these anatomical landmarks. If the markers will be placed on the skin, prior to marker placement, the skin will be cleaned with an alcohol swab and dried. Detailed instructions and a short practice period will be given to you before data collection starts. Overall we anticipate the entire experiment to take about an hour.

Spotters will be present during testing to monitor your safety. If at anytime you feel uncomfortable, simply inform the researcher and the assessment will stop immediately. You may also feel free to ask questions before or after data collection.

Your information and confidentiality:
Your identity will be protected. You will be assigned an ID number and any data collected will only be identified by that ID number only. We will collect two types of information from you. One is your personal identification information, the other is the

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experimental data. Your personal information (name and contact information) will only be used to contact you during this study. This information will not be used in the data analysis, nor will it be released to others. Personal information collected will be kept separate in a locked cabinet. Your identity will be kept confidential to the extent required by law.

It is important to point out that this research is basic research designed to help us understand the association between spinal cord injury and seated postural control. In any report stemming from this research only group data will be reported. No individual will be identified in any reports or presentation. The results of this study will inform us about how spinal cord injury impacts seated postural control. If you wish to have a copy of the results of this study please contact us.

Time requirements and risks:

The entire testing period should be completed in about an hour. We do not anticipate any physical risks greater than normal risks associated with daily life. There will always be at least two investigators nearby to monitor your safety. You will be given breaks throughout the data collection period as needed.

Refusal or withdrawal of participation:

Your participation is completely voluntary and you may withdraw participation at any time without penalty or loss of benefits to which you are otherwise entitled, and without influencing your relationship to the University of Illinois in any manner.

If you have any questions concerning this study please feel free to contact Dr. Jacob Sosnoff (email: jsosnoff@illinois.edu or phone: 217-333-9472). We appreciate your cooperation in this research. You will be given a copy of this consent form. For any questions regarding the rights of a research subject, please contact the Institutional Review Board Office of the University of Illinois at Urbana-Champaign at 217-333-2670 or irb@illinois.edu. Collect calls are welcome if you live outside of the local calling area.

Sincerely,

Jacob J. Sosnoff, Ph.D.
Assistant Professor
Department of Kinesiology and Community Health

Date_____

Project title: Seated postural control in individuals with spinal cord injury (SCI)

Agreement:

I have read and understand the procedures described above. I voluntarily agree to participate in this study and have been given a copy of this document to keep for my records. All of my questions about participating in this study have been answered satisfactorily. I understand that I may stop participating in this study at any time without penalty or loss of benefits. I do not waive any legal rights by signing this form.

Print name:_____

I am 18 years of age or older

Signature of participant:_____Date:_____

Appendix B

Testing procedures

This study functions out of a folder known as "SPC SCI." in the desktop of gait and balance lab. Researchers have to protect the identity of the participants. Please note that we have to be gentle and courteous with all the equipment. Keeps the lab space (including the shared computers desktop) clean and orderly. Do not use or move other peoples equipment without permission. Please also note that this study is researching a special population of wheelchair users. Please follow proper etiquette when interacting with wheelchair users. Consider their wheelchair part of their body, and ask permission before helping the subject transfer. Never force your help onto the subject, but always watch out for their personal safety.

Last edited by Sunghoon Shin and Shawna Culp, 8/29/2011

Questions? Contact jsosnoff@illinois.edu.

PROTOCOL:

Two days before data collection:

- Email the recruited subject using the "pre-testing Email Draft.doc" format. Use the subject: REMINDER – Research participation volunteer on (Date) at (Time)
- Ensure that all paperwork necessary is at the facility (print copies if necessary, they are in the folder "protocols and other paperwork" within the studys folder. This includes:
  - IRB consent form "IRB Consent form.doc"
  - Data sheet "Data Collection Form.doc"

  - Demographics survey "Wheelchair use demographics survey.doc"
  - Those materials should be prepared by Sunghoon before the actual data collection.

- Finishing calibration before using the piezosensor and force plate at least 15 min before data collection.

- Identify the subject number based on the private document. Assign a number and add _SPC1. (eg 01_SPC). This document details proper naming of the trials later on.

On the day of data collection 15-30 min before the subject arrives:

- Calibrate the Forceplate.
  - Open "Netforce" software of desktop for force plate.
  - Check to ensure that the "Hardware installation" is force plate 3057 in mini amplifier.
  - Check to see that the mini-amplifier connect to the computer and force plate.
  - Check to see that if the mini-amplifier is powered on.
  - Check to see that whether the time duration is 30 second and sampling rate is 100 Hz.

Once the subject arrives:

- Greet the subject at the front door and escort to the lab.
- Explain the procedures, and answer any questions.
- FIRST THING: HAVE THEM SIGN THE INFORMED CONSENT FORM!!!
- Ask the subject to use the facilities to help prevent them from leaving the chair during data collection.

- Ask the subject to fill out the wheelchair user survey form.
- Complete all measurements as outlined in the Data Collection form.
  - Anthropomorphic measurement procedure is outlined in the document "ANTHROPOMETRIC MEASUREMENTS PROTOCOL2.doc"
  - Please clean the measuring tape before/after each subject. Armpits are icky.
- Ask the subjects to remove all steel items to reduce artifact- men will be asked to be shirtless and women in a sports bra, unless it makes them uncomfortable. Tank tops are not ideal, but acceptable).
- Have the subject sit into the desk.

  Note: if the self-selected is the same as either fast or slow trial, we will not perform this trial.

Static condition

- Ask the subject to sit with thighs at 90 degrees and knees at 90 degrees. The subjects arms should be their side. There will be tape mark on the force plate. Center of subject should be lined up with marked tape on the center of force plate. It is very important to evaluate the asymmetry of COP.
- If researcher says "Ready! Set! Go!" and then the subject has to "subject hit the side of force plate with their hands on which magnetic sensor will be strapped to make synchronized point of each trials.
- For 30 seconds, they have to keep still.
- After 30 seconds the researcher has to say "Stop" and then subject will be relaxed with arm supporting on the any place around him or her.

- In supporting condition, subjects knee angle should be 90 degrees exactly. Support should check it before actual trial start.
- The initial supporting condition was randomized between subjects.

For each trial use this script: We are now going to complete static balance task. This task consists of eight 30-second trials.

During the each trial, if at any time you feel uncomfortable or in pain, please stop. we will ask you to stop and we will have to re-start the trial. If I say "move on the plate "And then sit on the force plate and follow direction. If I say "Ready! Set ! Go!" You should try to maintain quite still on the force plate. After 30 second, if I say "Stop" you will move out from the plate. After then for 30s you will be relaxed your can support your upper body.

If I say "Ready! Set ! Go!" You should try to maintain quite still on the force plate again.

During this trial, please sit with your hands on your side and stare at the red dot silently,

"Move on the Plate Please"

"Ready! Set! Go!"

After 30 seconds

"Stop"

30-second rest

This time you will do same thing **with your eyes closed**.

"Move on the Plate Please"

"Ready! Set! go!"

After 30 seconds

"Stop"

30-second rest

This time you will do same thing **with your eyes open.**

"Move on the Plate Please"

"Ready! Set! Go!"

After 30 seconds

"Rest"

After 30 seconds

This time you will do same thing **with your eyes closed.**

"Move on the Plate Please"

"Ready! Set! Go!"

After 30 seconds

"Rest"

After 30 seconds

This time you will do same thing **with your eyes open, but you will rest your feet on this foot support (place foot support under subjects feet).**

"Move on the Plate Please"

"Ready! Set! Go!"

After 30 seconds

"Rest"

This time you will do same thing **with your eyes closed with foot support.**

"Move on the Plate Please"

"Ready! Set! Go!"

After 30 seconds

"Rest"

This time you will do same thing **with your eyes open with foot support.**

"Move on the Plate Please"

"Ready! Set! Go!"

After 30 seconds

"Stop"

This time you will do same thing **with your eyes closed with foot support.**

"Move on the Plate Please"

"Ready! Set! Go!"

After 30 seconds

"Stop"

"We finished this session"

Save data using filenames of the form `(subject#)_Static`.

Functional boundary

- Before the actual trial starts, the researcher has to explain the task.
- Each subject will have them lean forward, backward, laterally and diagonally pivoting at the hip joints, in the circular direction leaning as far as possible without losing balance for one minute.
- If researcher says "Ready! Set! Go!" subjects pivot at the hip joints at their preferred pace without losing balance.
- Each trial will be having same procedure 1) without foot support, 2) with foot support.
- We are not allowed to talk to you during the trial, but we will inform you of your progress.

For each trial use this script: When we ask you to do each trial, please repeatedly hit the force plate with the arm strapped magnetic unit. If I say "Ready! Set! Go!", you start to stare at one focal point of well in front of you and try to keep quite still as much as possible.

If I say "stop" stop the task.

Move on the plate first.

For this trial, please keep your hands on your sides and your lower body still. Please rotate your upper body in its maximum range of motion in a circle while maintaining a motionless lower body. There will be no foot support for this trial. we will do this two times

"Ready! Set! go!"

After 30 seconds

"Stop"

"Move out please"

After 30 seconds

"We will do same thing one more time"

"Move on the Plate Please"

"Ready! Set! Go!"

After 30 seconds

"Stop"

We finished this session.

Save data using filenames of the form VTC boundary `(subject#)_boundary`.

Dynamic condition

- Participants will perform rhythmical rocking in both anterioposterior (AP) and mediolateral (ML) direction for 30s at 50 beat/min pace (0.83 Hz) speed.
- Practice at 1Hz rhythmical rocking:

  - Say “Now rock you body at a given pace with your low body as much as still.” For 30 second. Reminding and modifying their rocking not to move hip too much just rhythmically move their upper body at 50 beats/min.

- Actual trial at 1Hz rhythmical rocking:

  - Ask the subject to sit in a sitting position, the subjects arms should be on their sides comfortably. Center of subject should be lined up with marked tape on the center of force plate. It is very important to evaluate the asymmetry of COP.
  - During rocking they have not to use arm supporting.
  - If researcher says “Ready! Set! Go!”And then subject rock their body for 30 second. Researcher has to ask stare at focal point on the wall.

For each trial use this script: If I say "ready! Set! Go!", you start to stare at one focal point of well in front of you and try to rock your body for 30 second rhythmically.
Move on the plate first.

For this trial, please keep your hands on your sides and your lower body stationary. When I say go, rock forward and backwards as much as regularly in your preferred pace while keeping your lower body still. I will give the practice before actual trial (check their movement for 20 second if they understand the direction, start the actual trial).
"Ready! Set! Go!"

After 30 seconds

"Stop"

Researcher turns on the beat sound.

For this trial, please keep your hands on your sides and your lower body stationary. When I say go, rock forward and backwards as much as possible regularly in this beat sound while keeping your lower body still.
"Ready! Set! Go!"
After 30 seconds
"Stop"

Touching condition

- Participants will perform quiet sitting with touching support.
- Ask the subject to sit in a sitting position, the subjects arms should be on their thigh comfortably. There will be tape mark on the force plate. Center of subject should be lined up with marked tape on the center of force plate.
- In their right side there will be Piezo-force sensors whose will be monitored by one oscilloscope.
- Before the actual trial, height of force transducer should be adjusted so that the par-

ticipant's elbow will be 90 degrees.

- Before starting the light touching condition, researcher has the subjects learn what a force smaller than 1N feels like with their index finger touch the sensor.
- If researcher says "Ready! Set! Go!" And then subject has to sit still for 30 seconds with their right hand touching on the sensor. Researcher has to ask the subject to stare at focal point on the wall and screen whether their touching force is over 1N. If subjects touching force is over 1N, researcher should declare "discard the trial" and redo the trial.
- Before starting the heavy touching condition, researcher has the subjects learn what a force greater than 1N feels like with their index finger touch the sensor. Researcher has to explain that subject can use finger force as much as possible to give their body minimal sway.
- If researcher says "Ready! Set! Go!" And then the subject rocks their body for 1 minute. Researcher has to ask the subject to stare at focal point on the wall.

For each trial use this script: "We are now going to complete another 6 minute trial. If I say "Ready! Set! Go!" And then sit on the force plate and follow direction.

This task is consisting of 10 different 30 second trials. If I say "move on the plate"
And then sit on the force plate and follow direction. After 30 seconds, for 30s you will be relaxed your can support your upper body.
Two kinds of different touch will be used to help your balancing.
This is the light touching condition (actually touch the sensor and show the change of signal under two scales on the display of oscilloscope. You can touch this piezo sensor in front of your right hands. During the trial, keep your elbow in close to your body and keep 90 degree like this (actually show the posture).

This is heavy touch You can use touching force over 1N as much as possible to reduce your sway during the sitting (actually touch the sensor and show the change of signal over two scales on the display of oscilloscope).

For this trial, please sit quietly with your right finger in the sensor. Keep your force under 1N.
"Ready! Set! Go!"
After 30 seconds
"Stop"

30 second rest

For this trial we will do the same thing, however you may relay on the sensor as much as necessary to reduce your sway
"Ready! Set! go!"
After 30 seconds
"Stop"

30-second rest

For this trial, please sit on the dynadisk to reduce your stability. Your finger will lightly touch the sensor, however please maintain a force below 1N

“Ready! Set! Go!”

After 30 seconds

“Stop”

After 30 seconds

This time you will do same thing, however you can rely on the sensor as much as necessary to maintain your posture. Please maintain a force on the sensor of over 1N “Ready! Set! Go!”

After 30 seconds

“Stop”

After 30 seconds

For this trial, please sit on the dynadisk quietly with both hands on your hips. There will be no foot support for this trial.

“Ready! Set! Go!”

After 30 seconds

“Stop”

Repeat one more time.

After total 10 trials, “We finished this session”

Save data using filenames of the form `(subject#)_touch`.

Once all trials are completed:

Help the subject safely transfer out of force plate if they want. At least the supporter should keep an eye on and prepare to support their transfer.

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